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(71) Anmelder:
BrainLAB AG, 85551 Kirchheim, DE

(74) Vertreter:
Schwabe, Sandmair, Marx, 81677 München

(72) Erfinder:
Fröhlich, Stephan, 85609 Aschheim, DE;
Schlossbauer, Cornel, 82152 Krailling, DE;
Blumhofer, Andreas, Dr., 81677 München, DE

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(54) Exakte Patientenpositionierung durch Vergleich von rekonstruierten und Linac-Röntgenbildern

(57) Die Erfindung betrifft ein Verfahren zur exakten Positionierung eines Patienten für die Strahlentherapie bzw. Radiochirurgie mit folgenden Schritten: a) der Patient wird gegenüber einem Linearbeschleuniger vorpositioniert, b) mindestens eine Röntgenaufnahme des Patienten bzw. eines seiner Körperteile in der Umgebung des Bestrahlungszielpunktes wird erstellt, c) die Röntgenaufnahme wird lokalisiert, d) mindestens ein der Röntgenaufnahme, insbesondere isozentrisch, entsprechendes, aus einem Patientenscan stammendes rekonstruiertes Bild wird erstellt, e) das rekonstruierte Bild und die Röntgenaufnahme werden überlagert und anhand bestimmter Landmarken in beiden Bildern wird der Positionsfehler elektronisch bzw. computergesteuert ermittelt, und f) für Lage des Patienten wird anhand des ermittelten Positionierungsfehlers korrigiert. Ferner betrifft sie ein Verfahren zur räumlichen Lokalisation eines Röntgenbildes, bei dem ein Röntgenbild eines Patienten erstellt wird, bei der Erstellung des Röntgenbides die Raumlage des Röntgengeräts ermittelt wird, bei der Erstellung des Röntgenbildes Markierungen in einer vorbestimmten oder bekannten Lage gegenüber der Röntgenquelle in deren Strahlbereich eingebracht werden, und bei dem aus der Geometrie des Röntgengeräts und aus der Lage der Markierungen im Röntgenbild die exakte räumliche Aufnahmesituation des Röntgenbildes errechnet wird.

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Beschreibung

Die vorliegende Erfindung betrifft ein Verfahren zur exakten Positionierung eines Patienten für die Strahlentherapie bzw. Radiochirurgie. Ferner betrifft die vorliegende Erfindung ein Verfahren zur räumlichen Lokalisation eines Röntgenbildes.

In der Strahlentherapie und der Radiochirurgie wurden in letzter Zeit große Fortschritte in der Dosisplanung erzielt. Es wird angestrebt, die Behandlung immer weiter in Richtung der Radiochirurgie zu verlegen, d. h. mit hohen Strahlendosen zu arbeiten, die bei wenigen und vorzugsweise nur bei einer einzigen Strahlenbehandlung auf ein Zielvolumen aufgebracht werden, also beispielsweise auf einen Tumor. Obwohl die Dosisplanung, wie erwähnt, relativ gute Erfolge zeigt, steht der Verwendung von hohen Dosen, die in einer einzigen oder in wenigen Fraktionen verabreicht werden, oftmals die Tatsache im Wege, daß der Patient bzw. der zu bestrahlende Körperabschnitt, nur relativ ungenau positioniert werden kann. Um größere Schädigungen des gesunden Gewebes zu vermeiden, wird deshalb in den überwiegenden Fällen auf die konventionelle fraktionierte Strahlentherapie zurückgegriffen werden, bei der eine wiederholte Bestrahlung mit geringen Dosen appliziert wird.

Um die Positionierung zu verbessern behilft man sich derzeit noch eines sehr ungenauen "manuellen" Verfahrens. Dabei wird am Linearbeschleuniger ein Röntgenbild eines Patienten-Körperabschnittes erstellt. Dieses Bild wird mit einem zuvor am Simulator (Röntgengerät mit identischer Geometrie wie der Linearbeschleuniger) aufgenommenen Referenz-Röntgenbild verglichen. Nunmehr wird von dem behandelnden Arzt ein Vergleich des Röntgenbildes und des Simulator-Bildes beispielsweise an einem Lichtkasten vorgenommen. Dabei wird mit einem Lineal der Positionierungsfehler zwischen der tatsächlichen Patientenlage und der Soll-Lage ermittelt und der Patient wird daraufhin entsprechend verschoben. Bestenfalls steht dem Arzt noch ein Zentralstrahl-Kreuz und/oder die Kontour der äusseren Feldgrenzen in beiden Bildern als Anhaltspunkt zur Verfügung. Die Feldgrenzen könne z. B. durch Bleiblöcke bzw. verfahrbare Strahlblenden definiert sein. Auch beim Vergleich mit DDRs (virtuelle aus einem dreidimemsionalen Bilddatensatz ermittelte "Simulatorbilder") statt mit wirklichen Simulatorbildern ändert sich dieses Verfahren nicht.

Nachteiligerweise ist diese Art der Patientenpositionierung schon aus den folgenden Gründen ungenau:

Die Bilder sind projektiv und daher nicht im Originalmaßstab. (Es existiert kein einheitlicher Abbildungsmaßstab.)

Das "manuelle" Ablesen der benötigten Verschiebung ist ungenau.

Eine dreidimensionale räumliche Verschiebung ist aus zweidimensionalen Bildern ohne Computerunterstützung nur bedingt möglich und erfordert viel Erfahrung des Benutzers.

Aus der US-A 5,901,199 ist ein iteratives Verfahren zum Ausrichten von Therapie-Strahlen auf ein Behandlungsziel bekannt. Hierbei werden diagnostische Computertomographiedaten verwendet, mit Hilfe derer eine Vielzahl von rekonstruierten Röntgenbildern, so genannten DRRs (Digitally Reconstructed Radiographs) erzeugt werden. Diese DRRs werden so lange erstellt und immer wieder mit einem vor Ort aufgenommenen Röntgenbild verglichen, bis eines gefunden ist, welches eine ausreichende Übereinstimmung zeigt. Mit Hilfe der dabei erhaltenen Daten wird dann die Position des Behandlungsgerätes bzw. Behandlungsstrahls so korrigiert, daß der Strahl das Behandlungsziel trifft.

Nachteilig bei diesem Verfahren ist der hohe Rechenaufwand, da anfangs wahllos solche DRRs erzeugt werden müssen und der Vergleich sehr vieler DRRs mit dem tatsächlichen Röntgenbild vorgenommen werden muß. Insbesondere bedarf es der Auffindung eines "intelligenten" Algorithmus, um sich für jeden Körperabschnitt und für alle Patienten geltend in überschaubarer Zeit dem passenden DRR anzunähern.

Es ist die Aufgabe der vorliegenden Erfindung, ein Verfahren zur exakten Positionierung eines Patienten für die Strahlentherapie bzw. Radiochirurgie vorzuschlagen, bei dem die obigen Nachteile des Standes der Technik nicht mehr auftreten. Insbesondere soll eine sehr genaue Repositionierung des Patienten in einfacher Weise und in kurzer Zeit auf möglichst automatischem Wege erzielt werden. Diese Aufgabe wird erfindungsgemäß gelöst durch ein Verfahren zur exakten Positionierung eines Patienten für die Strahlentherapie bzw. Radiochirurgie mit den folgenden Schritten:

a) Der Patient wird gegenüber einem Linearbeschleuniger vorpositioniert,
b) mindestens eine Röntgenaufnahme des Patienten bzw. eines seiner Körperteile in der Umgebung des Bestrahlungszielpunktes wird erstellt,
c) die Röntgenaufnahme wird lokalisiert,
d) mindestens ein der Röntgenaufnahme insbesondere isozentrisch entsprechendes, aus einem dreidimensionalen Patientendatensatz rekonstruiertes Bild wird erstellt,
e) das rekonstruierte Bild und die Röntgenaufnahme werden überlagert und anhand bestimmter Landmarken in beiden Bildern wird der Positionsfehler elektronisch bzw. computergesteuert ermittelt, und
f) die Lage des Patienten wird anhand des ermittelten Positionierungsfehlers korrigiert.

Vorteilhafterweise bietet eine solche erfindungsgemäß vorgeschlagene Repositionierung einen relativ schnellen Weg, zu einer sehr exakten Zielbestrahlung zu gelangen. Die elektronische bzw. computergesteuerte Ermittlung des Positionsfehlers erhöht die Genauigkeit gegenüber dem "manuellen" Verfahren erheblich. Die Lokalisation der Röntgenaufnahme gestattet es, schon diesen Eingangswert mit ausreichender Genauigkeit in die Auswertung aufzunehmen, so daß auch von dieser Seite her Fehler und Verzögerungen bei der Repositionierung vermieden werden.

Bevorzugt erfolgt bei einem erfindungsgemäßen Verfahren die Vorpositionierung mittels eines computer- und kameragesteuerten Navigations- und Trackingsystems mit Hilfe künstlicher, insbesondere reflektierender Markeranordnungen an dem Patienten und den Behandlungseinrichtungen. Ein solches Navigations- und Trackingsystem kann alle notwendigen Positionserfassungen bei der Durchführung des erfindungsgemäßen Verfahrens übernehmen und entsprechende Informationen beispielsweise auf einem Computerbildschirm ausgeben.

Die Vorpositionierung kann aber auch über Hautmarkierungen am Patienten, über natürliche Landmarken oder Lasermarkierungen erfolgen.

Es sollte grundsätzlich genügen, lediglich eine Röntgenaufnahme und ein entsprechend rekonstruiertes Bild zu erstellen. Bei bevorzugten Ausführungsformen des erfindungsgemäßen Verfahrens werden jedoch mindestens zwei oder mehrere Röntgenaufnahmen und entsprechend viele rekonstruierte Bilder aus verschiedenen Richtungen erstellt und jeweils durch Vergleich ausgewertet, um auch Verkippungen des Patienten bzw. des Patiententrägers mit einberechnen zu können.

Die Röntgenaufnahme kann vorteilhafterweise mit Hilfe des Linearbeschleunigers erstellt werden. Solche Röntgenbilder werden EPID-Bilder (Electronic Portal Imaging Device-Bilder) genannt und die entsprechenden Aufnahmen können auf einem Flat Panel (z. B. amorphes Silizium) auf einem Röntgenfilm oder auf jedem anderen zweidimensionalen Bildaufnehmer erstellt werden.

Andererseits ist es natürlich auch möglich, die Röntgenaufnahmen durch eine separate Röntgenquelle zu erzeugen. Dies kann z. B. mit Hilfe von zwei an der Decke befestigten Röntgenquellen geschehen, die sequenziell (elektronische) Röntgenbilder auf einem Detektor (z. B. amorphes Silizium) erzeugen. Falls der Detektor aus diversen Gründen (z. B. Rotation der Gantry) nicht im Isozentrum positioniert werden kann, muß ein Offset sowohl bei der Grobpositionierung, als auch bei der Fehlerkorrektur berücksichtigt werden.

Ganz allgemein kann die Röntgenaufnahme auf einem Bildverstärker oder Detektor, insbesondere auf dem genannten amorphen Silizium erzeugt werden, wobei bei der Verwendung von amorphem Silizium (Flat Panel) Verzerrungen minimiert werden. Natürlich ist auch die Verwendung eines eingescannten Röntgenfilmes möglich. Die Röntgenaufnahme kann sowohl durch ein in den Linearbeschleuniger integriertes Bildsystem als auch durch ein separates Röntgengerät erzeugt werden.

Bei einer Ausführungsform des erfindungsgemäßen Verfahrens erfolgt die Überlagerung der Röntgenaufnahme und des rekonstruierten Bildes durch ein vom Benutzer gesteuertes Markieren und Übereinanderschieben auf einem Computerbildschirm (z. B. mit Maus, Keyboard, Touchscreen, Joystick, etc.). Andererseits kann die Überlagerung der Röntgenaufnahme und des rekonstruierten Bildes natürlich auch durch eine rechnergesteuerte automatische Bildfusion erfolgen.

Bei bevorzugten Ausführungsformen des erfindungsgemäßen Verfahrens werden das rekonstruierte Bild bzw. die rekonstruierten Bilder erstellt als:

- Digitally Reconstructed Radiographs (DRRs),
- Digitally Composited Radiographs (DCRs)
- MIP-Images,

oder als jedwede zweidimensionale Bildrekonstruktion aus einem dreidimensionalen Patientenscan-Datensatz.

Die Lage des Patienten wird erfindungsgemäß vorteilhafterweise durch die Verschiebung des Patiententisches korrigiert, insbesondere automatisch angesteuert und korrigiert durch ein computer- und kameragesteuertes Navigations- und Trackingsystem mit Markern am Patienten und/oder an dem Patiententisch. Grundsätzlich besteht daneben auch die Möglichkeit, die Lage des Patienten durch eine manuelle Tischansteuerung zu korrigieren.

Gemäß einer bevorzugten Ausführungsform des erfindungsgemäßen Verfahrens werden in den oben angeführten Schritten c) und d) eine Vielzahl von rekonstruierten Bildern erstellt und elektronisch bzw. computergesteuert mit der lokalisierten Röntgenaufnahme überlagert und verglichen, bis ein der Röntgenaufnahme entsprechendes rekonstruiertes Bild gefunden ist, anhand dessen dann der Positionsfehler ermittelt wird.

Man benötigt hier keine isozentrisch entsprechenden, rekonstruierten Bilder, weil man sich durch rechnerische Annäherungsverfahren (Algorithmen) immer weiter dem gesuchten rekonstruierten Bild nähern kann. Besonders vorteilhaft ist diese Ausführungsform deshalb, weil sie einen breiteren Spielraum bei der Vorpositionierung gestattet. Durch die Verwendung eines lokalisierten Röntgenbildes lässt sich ein schnelleres und genaueres Auffinden des entsprechenden rekonstruierten Bildes realisieren.

Die Erfindung betrifft ferner ein Verfahren zur räumlichen Lokalisation eines Röntgenbildes, bei dem:

- ein Röntgenbild eines Patienten erstellt wird,
- bei der Erstellung des Röntgenbildes die Raumlage des Röntgengerätes ermittelt wird,
- bei der Erstellung des Röntgenbildes Markierungen in einer vorbestimmten bzw. bekannten Lage gegenüber der Röntgenquelle in deren Strahlbereich eingebracht werden, und bei dem
- aus der Geometrie des Röntgengerätes und aus der Lage der Markierungen im Röntgenbild die exakte räumliche Aufnahmesituation des Röntgenbildes errechnet wird.

Durch das obige erfindungsgemäße Verfahren läßt sich nunmehr die räumliche Lage eines Röntgenbildes exakt ermitteln. Dies ist insbesondere dann wichtig, wenn dieses Röntgenbild als ein Eingangsparameter für weitere Lokalisierungen und Positionierungen verwendet wird, da dadurch schon dieser Eingangswert genau lokalisiert und fehlerfrei angegeben werden kann. Bei der Bilderzeugung an einem Linearbeschleuniger ist oftmals die Position des Bildverstärkers bzw. des Films auf dessen Halterung nicht 100%ig fest gegenüber der Strahlungsquelle und gegenüber dem Isozentrumsstrahl. Durch die Lokalisierung jedes einzelnen Röntgenbildes kann ein solcher Fehler ausgeschlossen werden.

Es besteht hierbei die Möglichkeit, die Raumlage der Röntgenquelle und/oder des Bildempfängers sowie eines Patiententrägers mittels eines computer- und kameragesteuerten Navigations- und Trackingsystem mit Markern zu ermitteln. Ferner kann die Erfassung der Raumlage der Röntgenquelle und/oder des Bildempfängers aber auch über skalierte Erfassungseinrichtungen an diesen Geräten erfolgen.

Bei einer Ausführungsform des Lokalisationsverfahrens wird die Röntgenaufnahme durch einen Linearbeschleuniger für die Strahlentherapie bzw. Radiochirurgie mit einem Bildempfänger erstellt, wobei ein Träger für die Markierungen fest vor der Strahlungsquelle positioniert wird. Die Markierungen erscheinen auf dem Röntgenbild und wegen des bekannten Abstandes der Markierungen zur Strahlungsquelle sowie der bekannten Markierungsgeometrie ist es dann möglich, die räumliche Aufnahmesituation des Röntgenbildes exakt zu errechnen.

Vorteilhafterweise wird ein Linearbeschleuniger mit einem Lamellenkollimator vor der Strahlungsquelle verwendet, wobei die Markierungen durch bis zu einem bestimmten Grad in den Strahlengang eingefahrene Kollimatorlamellen gebildet werden. Das Feld des Lamellenkollimators kann dabei entweder bereits in der Bestrahlungsform sein, oder aber speziell zur Lokalisation geformt werden, wobei die Lamellen asymetrisch nur am Rand ausgefahren werden, um das Bild nicht zu beeinträchtigen. In der Regel sind die Abstände zwischen der Strahlungsquelle und dem Markierungsträger bzw. dem Lamellenkollimator fest und bekannt. Eventuell kann aber eine Kalibrierung mit einem Phantom noch genauere Werte bereitstellen.

Erfindungsgemäß ist es natürlich möglich und bevorzugt auch vorgesehen, das beschriebene Verfahren zur räumlichen Lokalisation eines Röntgenbildes dazu zu verwenden, eine Röntgenaufnahme im Rahmen des zuerst beschriebenen Verfahrens zur exakten Positionierung eines Patienten zu verwenden.

Die Erfindung wird im Weiteren anhand der beiliegenden Zeichnungen näher erläutert

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Zeichnungen näher erläutert. Es zeigen:

Fig. 1a und 1b zwei verschiedene Aufnahmesituationen für Röntgenbilder, die mit einem Linearbeschleuniger erzeugt werden;

Fig. 2 eine schematische Darstellung der Erzeugung zweier rekonstruierter Bilder entsprechend den Aufnahmen in den **Fig.** 1a und 1b;

Fig. 3 die Einbringung eines Markierungsträgers in einem Einschub am Linearbeschleuniger sowie zwei Beispiele für Markierungsgeometrien;

Fig. 4 Röntgenbilder, bei denen Kolimatorlamellen als Markierungen abgebildet sind;

Fig. 5 eine Überlagerung eines Röntgenbildes und eines rekonstruierten Bildes; und

Fig. 6 eine Prinzipdarstellung für die Herstellung von Röntgenaufnahmen mit separaten Röntgenquellen.

In den **Fig.** 1A und 1B wird gezeigt, wie aus zwei verschiedenen Richtungen Röntgenbilder mit Hilfe eines Linearbeschleunigers erstellt werden. Der Linearbeschleuniger weist eine Gantry **1** auf, die schwenkbar angeordnet und in deren Oberteil die Strahlungsquelle **2** angeordnet ist. Wie aus **Fig.** 1b hervorgeht, ist am Unterteil der Gantry **1** ein Träger **9** ausklappbar befestigt, der ein Bildsystem **5** aufweist, auf dem der Röntgenbildempfänger angeordnet ist. Ein solcher Bildempfänger kann ein einfacher Röntgenfilm sein (wie dargestellt), aber auch ein Flat Panel (z. B. amorphes Silizium) oder ein Bildverstärker. Der isozentrische Strahl **8** aus der Strahlungsquelle **2** geht durch den Patienten **7** hindurch auf das Bildsystem. Auf dem Röntgenfilm **6** wird dabei ein Röntgenbild erzeugt. Hierzu wird der Patient **7** mit Hilfe des Patiententisches **4**, der durch nicht dargestellte Motoren verfahrbar ist, schon so gut wie möglich vorpositioniert. Dies kann mit Hilfe eines bekannten Trackingsystems und an dem Patienten bzw. an dem Patiententisch angebrachten Markern geschehen.

Für die hier dargestellte Ausführungsform der Erfindung werden demnach, wie in den **Fig.** 1a und 1b gezeigt, zwei Röntgenbilder aus um etwa 90° verdrehten Winkeln der Gantry erstellt.

In **Fig.** 2 ist nunmehr schematisch dargestellt, wie zwei entsprechende, rekonstruierte Bilder erstellt werden. Hierzu wird ein Computertomographie-Scan-Datensatz verwendet, der vorher von dem Patienten erstellt wurde. In **Fig.** 2 ist dieser Datensatz durch eine hintereinander angeordnete Vielzahl von Schnittbildern **20** veranschaulicht. Mit den bekannten Positionsdaten der Strahlungsquelle **2** (siehe **Fig.** 1a und 1b) werden nun anhand der eingescannten Daten entsprechende rekonstruierte Röntgenbilder **16a** und **16b** erzeugt. Die isozentrischen Strahlen sind mit 18a und 18b bezeichnet.

Eingangsdaten für die Erzeugung der rekonstruierten Bilder, die im weiteren auch DRRs (Digitally Reconstructed Radiographs) genannt werden, sind zum einen die Positionen der Strahlungsquellen **12a** und **12b**. Als zweite Eingangsgröße muß dann die räumliche Anordnung derjenigen Ebene vorgegeben werden, in der das Röntgenbild erstellt wird, und zwar sowohl hinsichtlich des Abstandes zur Strahlungsquelle als auch hinsichtlich ihrer Verkippung. Mit anderen Worten müssen die "virtuellen" Röntgenfilme **16a** und **16b** genau in der gleichen Weise angeordnet sein, wie die Filme **6** aus der tatsächlichen Röntgenaufnahme, um die Bilder vergleichbar zu machen. Dazu müssen die Röntgenbilder **6** aus der tatsächlichen Röntgenaufnahme vor Ort (**Fig.** 1a und 1b) lokalisiert werden, d. h. hinsichtlich ihres Abstandes und ihrer Verkippung gegenüber der Strahlungsquelle genau bestimmt werden, wie dies detailliert später erläutert wird. Als dritte Eingangsgröße muß die Position des Zielpunktes im 3D-Datensatz bekannt sein.

Wenn dann die Röntgenbildebene und die Richtung des Zentralstrahls (Position der Strahlungsquelle) bei der tatsächlichen Röntgenaufnahme (**Fig.** 1a und 1b) vor Ort genau bekannt sind, können exakt die entsprechenden DRRs rekonstruiert und zugeordnet werden.

Anhand der **Fig.** 3 und 4 soll nunmehr beschrieben werden, wie das vor Ort erstellte Röntgenbild (**Fig.** 1a und 1b) hinsichtlich seines Abstandes und seiner Verkippung gegenüber der Strahlungsquelle lokalisiert wird.

Notwendig wird solch eine Lokalisierung vor allem deshalb, weil die Position des Bildsystems **5** sowie des ausklappbaren Trägers **9** gegenüber der Strahlungsquelle **2** aus mechanischen Gründen (Wackeln; ungenaue Klappmechanik) nicht als fest angesehen werden kann. Auch wenn sich diese Position nur um einige Millimeter ändert, kann dies zu unerwünschten Fehlbestrahlungen führen. Deshalb wird erfindungsgemäß bei jeder Erstellung eines Röntgenbildes dessen räumliche Anordnung bestimmt um dann das entsprechende DRR in exakt derselben Ebene rekonstruieren zu können.

Die obere Abbildung in **Fig.** 3 zeigt schematisch einen Einschub **30**, der mit Markern **33** versehen ist und welcher in einer hierfür bestimmten Halterung **10** an der Gantry **1** in den Strahlengang eingeschoben wird. Der Einschub **10** trägt entweder kreisförmige Marker **32** oder aber linienförmige Markierungen **31**, und in der erstellten Röntgenaufnahme bilden sich diese Marker ab, wie aus den beiden unteren Darstellungen in **Fig.** 3 ersichtlich wird. Aus dem Abstand und der Verzerrung der Geometrie der Marker **32**, bzw. der Linien **31** läßt sich dann mit einfachen geometrischen Mitteln bestimmen, in welchem Abstand das Röntgenbild aufgenommen wurde und wie es dabei verkippt war. Das vor Ort gemachte Röntgenbild ist damit genau lokalisiert und mit Hilfe dieser Daten lassen sich in exakt derselben Ebene DRRs erzeugen, wie dies unter Bezugnahme auf die **Fig.** 2 oben erläutert wurde.

Bei einer weiteren Ausführungsform des erfindungsgemäßen Verfahrens werden die zur Lokalisation erforderlichen Marker nicht durch einen zusätzlichen Einschub erzeugt, sondern durch einen Lamellenkolimator, welcher zur Ausformung des Behandlungsstrahles in vielen Fällen schon im Strahlengang des Linearbeschleunigers angeordnet ist. Solche Lamellenkolimatoren besitzen in den Strahlengang einfahrbare Lamellen und begrenzen mit diesen ein Strahlenfeld gemäß der Außenform der zu bestrahlenden Läsion, um dadurch umgebendes gesundes Gewebe zu schützen. Solche durch vorgeschobene Lamellen **33** eingegrenzten Bestrahlungsfelder sind in den beiden oberen Darstellungen der **Fig.** 3 zu sehen.

Es ist erfindungsgemäß aber auch möglich solche einfahrbaren Lamellen als Markierungen zur Lokalisation des Röntgenbildes zu verwenden. Hierzu werden, wie dies aus den beiden unteren Darstellungen in **Fig.** 4 hervorgeht, einzelne Lamellen **34** bei der Erstellung der Röntgenaufnahme in den Bildbereich eingeschoben. Auch von diesen Lamellen **34** sind Abstand und Anordnung gegenüber der Strahlungsquelle bekannt, so daß aus den Projektionen auf der Röntgenabbildung wiederum wie mit den separaten Markierungen **31** oder **32** (**Fig.** 4) die Raumlage, d. h. Verkippung und Abstand des erstellten Röntgenbildes ermittelt werden können. Um das Bild nicht zu sehr zu stören, werden hierzu in asymmetrischer Weise lediglich Lamellen in der Peripherie des Bildes eingefahren.

Weil das vor Ort erstellte Röntgenbild vollständig lokalisiert wurde und die jeweiligen DRRs in exakt derselben Ebene erstellt werden konnten, sind nunmehr die mit diesen beiden Verfahren erzeugten Bilder unmittelbar vergleichbar. Die **Fig.** 5 zeigt eine schematische Darstellung in der zwei

solche Bilder, nämlich das vor Ort erstellte Röntgenbild und das entsprechende DRR übereinandergelegt wurden. Der besseren Darstellbarkeit halber wurden als Bildobjekte hier Ringe gewählt, üblicherweise werden hier beispielsweise Knochenstrukturen zu sehen sein. In **Fig.** 5 wird ersichtlich, daß die beiden Abbildungen des Rings, nämlich die Abbildung aus der Röntgenaufnahme vor Ort, die mit dem Bezugszeichen **6** versehen ist, und die Abbildung aus dem DRR mit dem Bezugszeichen **16** zueinander verschoben sind. Diese Verschiebung resultiert aus der noch relativ ungenauen Vorpositionierung. Es wird nunmehr mit Hilfe einer rechnergesteuerten Bildverarbeitung (Fusion) oder manuell der Versatz der beiden Ringe zueinander in mindestens einer, vorzugsweise aber mehreren Ebenen (bei zwei oder mehreren Röntgenaufnahmen bzw. DRRs aus verschiedenen Richtungen) bestimmt und der Patient kann entsprechend dieses festgestellten Versatzes repositioniert werden. Dies kann bevorzugt automatisch über die Fahrmotoren des Patiententisches **4** (**Fig.** 1a oder 1b) erfolgen. Nach dieser Repositionierung befindet sich der Patient dann in exakt einer solchen Position, in der der Behandlungsstrahl das Behandlungsziel genau trifft. Die Behandlung kann durchgeführt werden.

Es ist anzumerken, daß die beiden Bilder, wenn sie übereinander gelegt werden, beispielsweise auch auf einem Computerbildschirm markiert manuell so übereinandergezogen gelegt werden können, bis die Abbildungen in allen Ebenen übereinstimmen. Die Bildinhalte, die aufeinandergelegt werden, werden bei Röntgenaufnahmen meist Knochenschatten sein. Es soll an dieser Stelle aber bemerkt werden, daß grundsätzlich auch andere Scanverfahren mit dem Prinzip der vorliegenden Erfindung verwendbar sind. Beispielsweise ist hier an die Verwendung von Kernspintomographie-Aufnahmen zu denken, welche dann relativ gute Abbilder des Behandlungsziels selbst ergeben, so daß der Positionsfehler durch eine Ermittlung des Versatzes des Behandlungsziels selbst in vorteilhafter Weise ermittelt werden kann.

In **Fig.** 6 ist noch eine alternative Ausführungsform dargestellt, bei der nicht der Linearbeschleuniger mit der Gantry **1** selbst zur Erzeugung des Röntgenbildes vor Ort verwendet wird, sondern vielmehr separate Röntgenquellen. Diese beiden separaten Röntgenquellen zur Erzeugung von zwei Röntgenbildern aus verschiedenen Richtungen sind in **Fig.** 6 nur schematisch dargestellt und mit den Bezugszeichen **51** und **52** versehen. Auch vor diese Röntgenquellen **51** und **52** können, wie anhand der **Fig.** 3 beschrieben, in vorbestimmtem Abstand Markierungen angebracht werden, so daß sich die exakte Lage des Röntgenbildes auf dem Bildsystem **5** immer lokalisieren läßt und der Patient auf seinem Patiententisch **4** nach der Bestimmung des Positionierungsfehlers exakt repositioniert werden kann. Vorteilhafterweise muß bei dieser Ausführungsform die Gantry zur Erstellung der Röntgenaufnahmen nicht bewegt werden.

Patentansprüche

1. Verfahren zur exakten Positionierung eines Patienten für die Strahlentherapie bzw. Radiochirurgie mit den folgenden Schritten:
   a) der Patient wird gegenüber einem Linearbeschleuniger vorpositioniert,
   b) mindestens eine Röntgenaufnahme des Patienten bzw. eines seiner Körperteile in der Umgebung des Bestrahlungszielpunktes wird erstellt,
   c) die Röntgenaufnahme wird lokalisiert,
   d) mindestens ein der Röntgenaufnahme, insbesondere isozentrisch, entsprechendes, aus einem dreidimensionalen Patientenscan-Datensatz stammendes rekonstruiertes Bild wird erstellt,
   e) das rekonstruierte Bild und die Röntgenaufnahme werden überlagert und anhand bestimmter Landmarken in beiden Bildern wird der Positionsfehler elektronisch bzw. computergesteuert ermittelt, und
   f) die Lage des Patienten wird anhand des ermittelten Positionierungsfehlers korrigiert.

2. Verfahren nach Anspruch 1, bei dem die Vorpositionierung mittels eines computer- und kameragesteuerten Navigations- und Trackingsystems mit Hilfe künstlicher, insbesondere reflektierender Markeranordnungen an dem Patienten und an den Behandlungseinrichtungen erfolgt.

3. Verfahren nach Anspruch 1, bei dem die Vorpositionierung über Hautmarkierungen an dem Patienten, über natürliche Landmarken oder Lasermarkierungen erfolgt.

4. Verfahren nach einem der Ansprüche 1 bis 3, bei dem mindestens zwei oder mehrere Röntgenaufnahmen und entsprechende rekonstruierte Bilder aus verschiedenen Richtungen erstellt und jeweils durch Vergleich ausgewertet werden.

5. Verfahren nach einem der Ansprüche 1 bis 4, bei dem die Röntgenaufnahme mit der Strahlenquelle des Linearbeschleunigers erstellt wird.

6. Verfahren nach einem der Ansprüche 1 bis 4, bei dem die Röntgenaufnahme durch eine, zwei oder mehrere separate Röntgenquellen erzeugt werden.

7. Verfahren nach Anspruch 5 oder 6, bei dem die Röntgenaufnahme auf einem Bildverstärker oder Detektor, insbesondere auf amorphem Silizium erzeugt wird.

8. Verfahren nach Anspruch 5 oder 6, bei dem die Röntgenaufnahme auf einem Röntgenfilm erzeugt wird.

9. Verfahren nach einem der Ansprüche 5 bis 8, bei dem die Röntgenaufnahme und/oder das rekonstruierte Bild auf einem Bildschirm ausgegeben werden.

10. Verfahren nach Anspruch 9, bei dem die Überlagerung der Röntgenaufnahme und des rekonstruierten Bildes durch ein manuelles Markieren und Übereinanderschieben auf einem Computerbildschirm erfolgt.

11. Verfahren nach einem der Ansprüche 1 bis 9, bei dem die Überlagerung der Röntgenaufnahme und des rekonstruierten Bildes durch eine rechnergesteuerte automatische Bildfusion erfolgt.

12. Verfahren einem der Ansprüche 1 bis 11, bei dem das rekonstruierte Bild bzw. die rekonstruierten Bilder erstellt werden als:
   - Digitally reconstructed Radiographs (DRRs),
   - Digitally Composited Radiographs (DCRs),
   - MIP-Images,

oder als jedwede zweidimensionale Bildrekonstruktion aus einem dreidimensionalen Patientenscan-Datensatz.

13. Verfahren nach einem der Ansprüche 1 bis 12, bei dem die Lage des Patienten durch die Verschiebung des Patiententisches korrigiert wird, insbesondere automatisch angesteuert und korrigiert durch ein computer- und kameragesteuerten Navigations- und Trackingsystem mit Markern am Patienten und an dem Patiententisch.

14. Verfahren nach einem der Ansprüche 1 bis 12, bei dem die Lage des Patienten durch eine manuelle Tischansteuerung korrigiert wird.

15. Verfahren nach Anspruch 1, bei dem in den Schritten c) und d) eine Vielzahl von rekonstruierten Bildern

erstellt und elektronisch bzw. computergesteuert mit der lokalisierten Röntgenaufnahme überlagert und verglichen werden, bis ein der Röntgenaufnahme entsprechendes rekonstruiertes Bild gefunden ist, anhand dessen dann der Positionsfehler ermittelt wird.

16. Verfahren zur räumlichen Lokalisation eines Röntgenbildes, bei dem:
- ein Röntgenbild eines Patienten erstellt wird,
- bei der Erstellung des Röntgenbildes die Raumlage des Röntgengeräts ermittelt wird,
- bei der Erstellung des Röntgenbildes Markierungen in einer vorbestimmten oder bekannten Lage gegenüber der Röntgenquelle in deren Strahlbereich eingebracht werden, und bei dem
- aus der Geometrie des Röntgengeräts und aus der Lage der Markierungen im Röntgenbild die exakte räumliche Aufnahmesituation des Röntgenbildes errechnet wird.

17. Verfahren nach Anspruch 16, bei dem die Raumlage der Röntgenquelle und/oder des Bildempfängers sowie eines Patiententrägers mittels eines computer- und kameragesteuerten Navigations- und Trackingsystems mit Markern ermittelt wird.

18. Verfahren nach Anspruch 16, bei dem Raumlage der Röntgenquelle und/oder des Bildempfängers und/oder des Patiententrägers über skalierte Erfassungseinrichtungen an diesen Geräten erfolgt.

19. Verfahren nach einem der Ansprüche 16 bis 18, bei dem die Röntgenaufnahme durch einen Linearbeschleuniger für die Strahlentherapie bzw. Radiochirurgie mit einem Bildempfänger erstellt wird, wobei ein Träger für die Markierungen fest vor der Strahlungsquelle positioniert wird.

20. Verfahren nach Anspruch 19, bei dem ein Linearbeschleuniger mit einem Lamellenkollimator vor der Strahlungsquelle verwendet wird, wobei die Markierungen durch bis zu einem bestimmten Grad in den Strahlengang eingefahrene Kollimatorlamellen gebildet werden.

21. Verfahren nach einem der Ansprüche 1 bis 15, bei dem zur Lokalisation der Röntgenaufnahme ein Verfahren gemäß den Ansprüchen 16 bis 20 verwendet wird.

Hierzu 4 Seite(n) Zeichnungen

Nummer: DE 199 53 177 A1
Int. Cl.[7]: A 61 B 6/04
Offenlegungstag: 21. Juni 2001

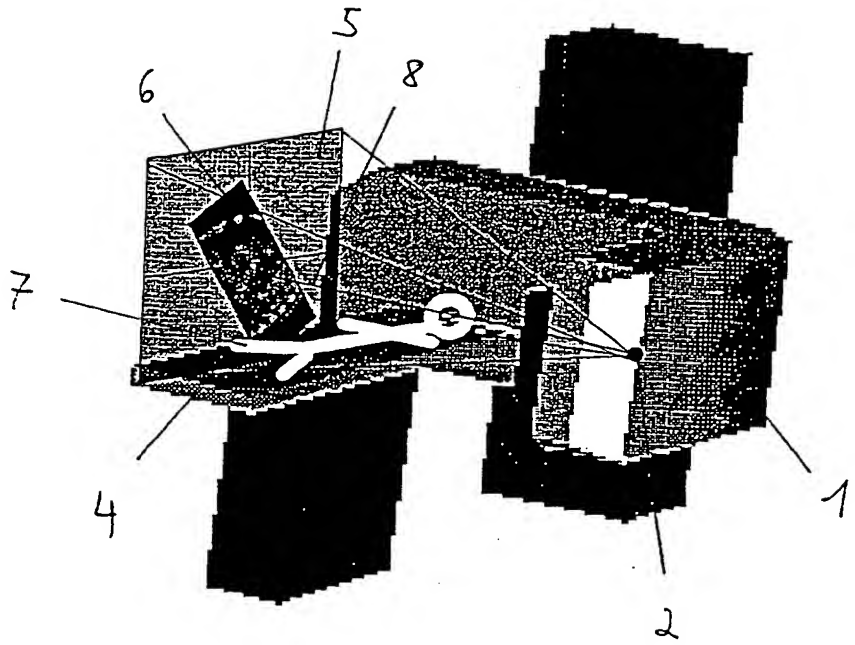


Fig. 1a

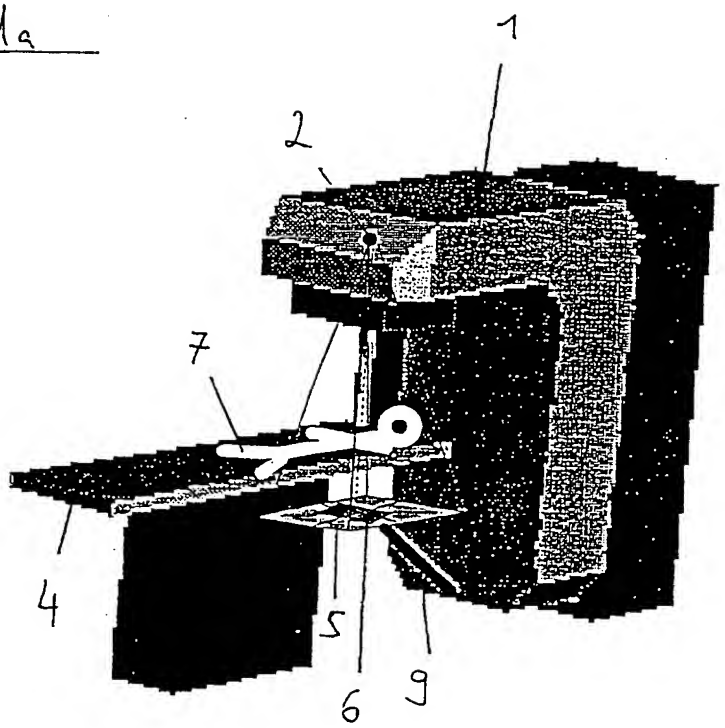


Fig. 1b

Nummer: DE 199 53 177 A1
Int.Cl.[7]: A 61 B 6/04
Offenlegungstag: 21. Juni 2001

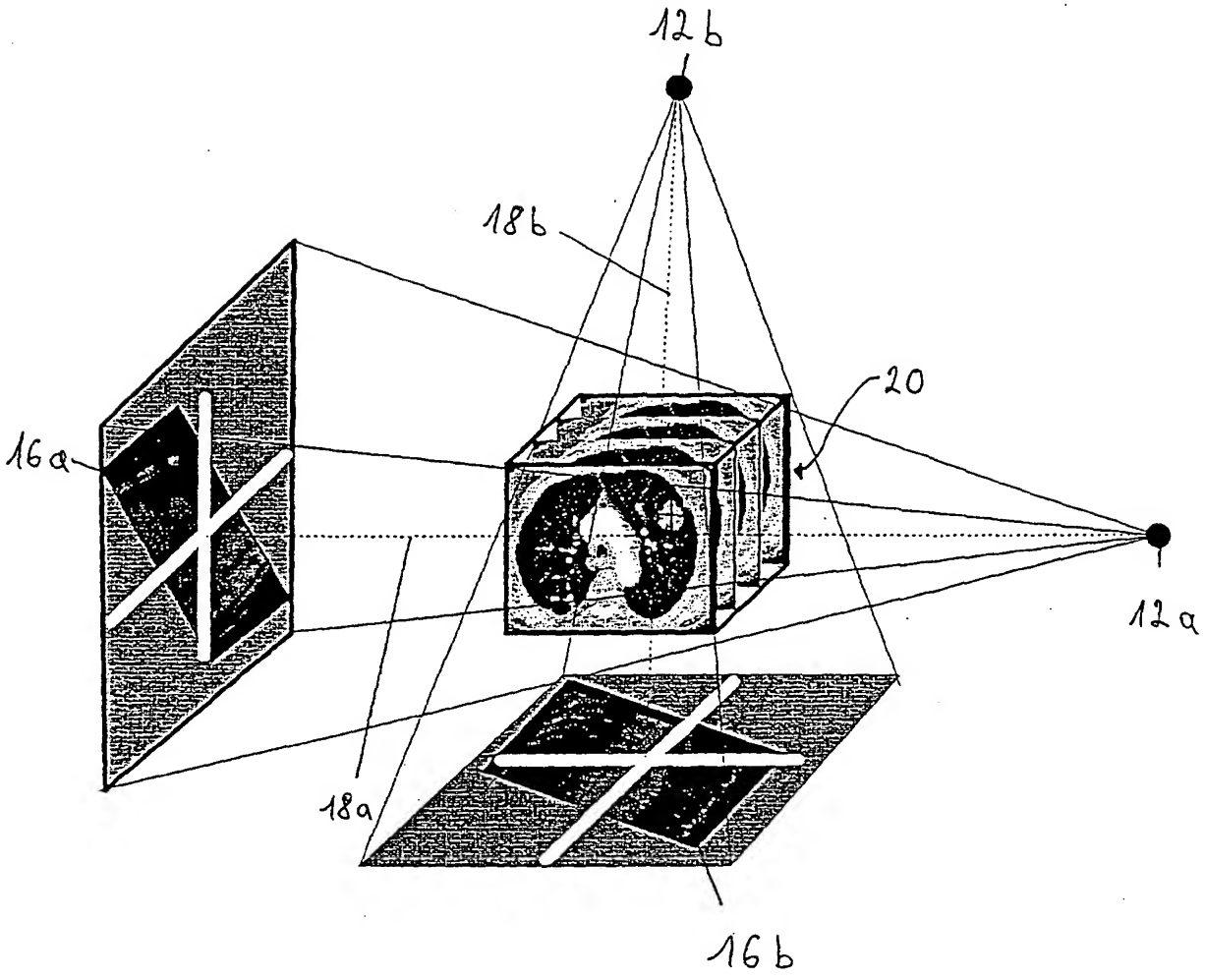
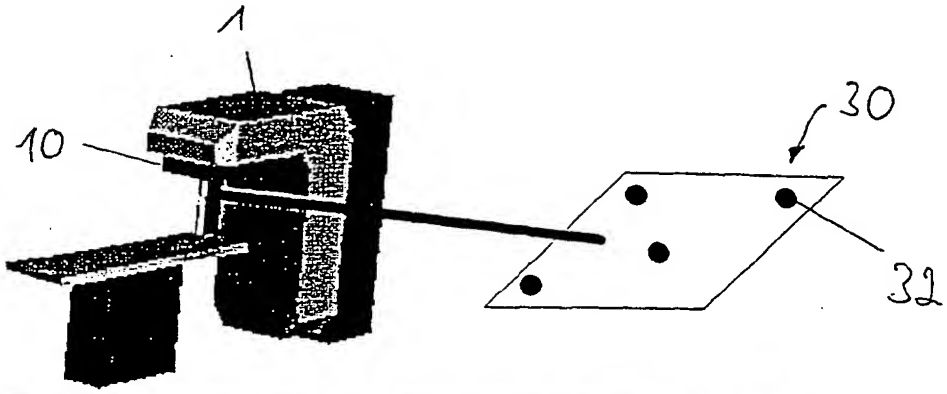


Fig. 2

Nummer: DE 199 53 177 A1
Int. Cl.[7]: A 61 B 6/04
Offenlegungstag: 21. Juni 2001



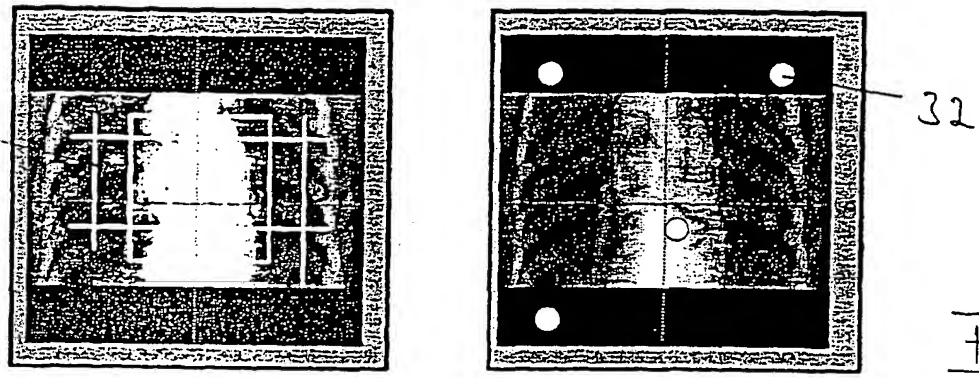
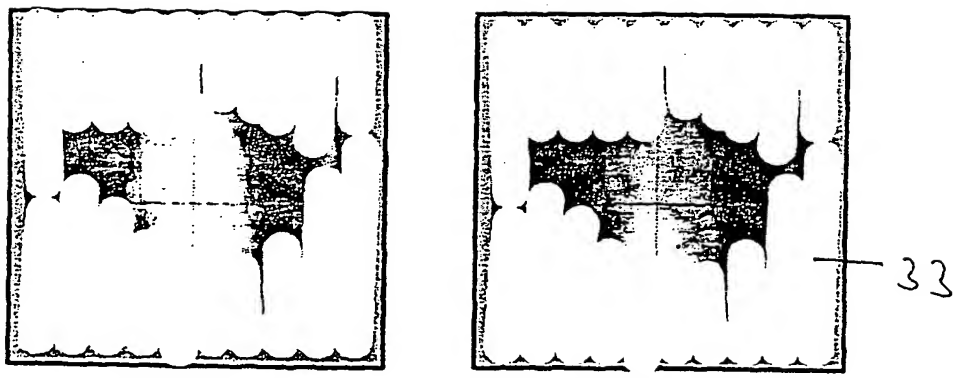


Fig. 3



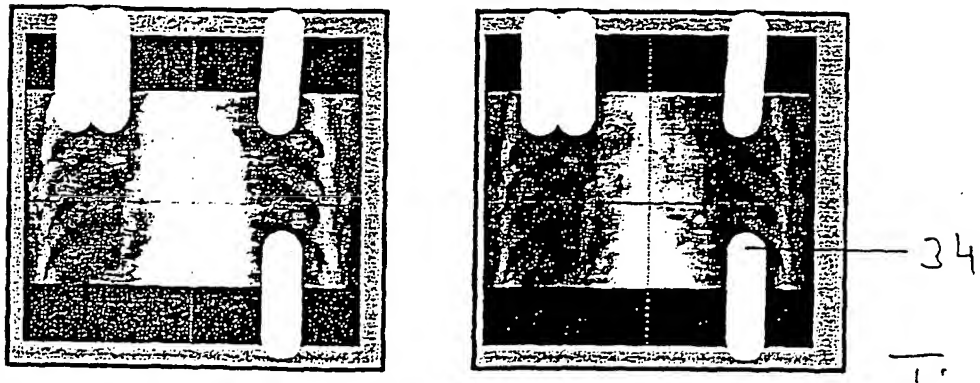


Fig. 4

Nummer: DE 199 53 177 A1
Int. Cl.[7]: A 61 B 6/04
Offenlegungstag: 21. Juni 2001

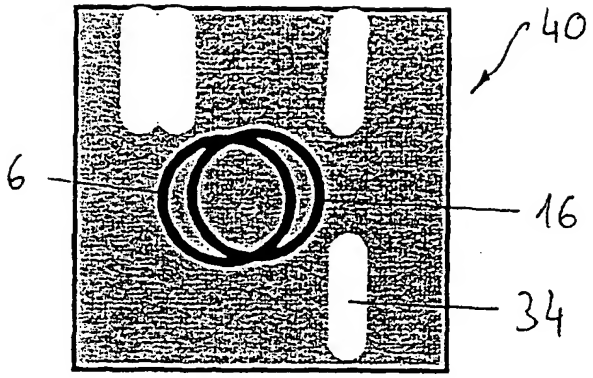


Fig. 5

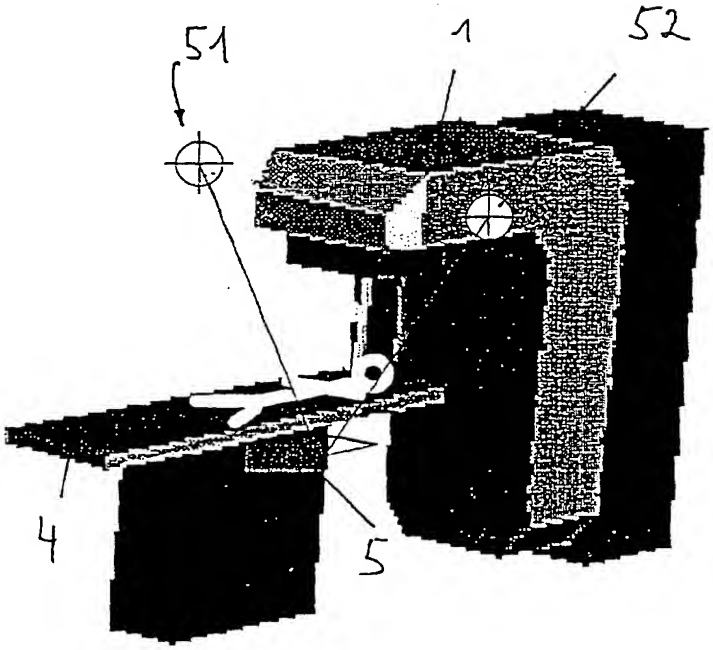


Fig. 6

US005394875A

United States Patent [19]

Lewis et al.

[11] Patent Number: 5,394,875

[45] Date of Patent: Mar. 7, 1995

[54] AUTOMATIC ULTRASONIC LOCALIZATION OF TARGETS IMPLANTED IN A PORTION OF THE ANATOMY

[76] Inventors: Judith T. Lewis, 855 Glendale La., Nashville, Tenn. 37204; Robert L. Galloway, Jr., 7736 Indian Springs Dr., Nashville, Tenn. 37221

[21] Appl. No.: 139,139

[22] Filed: Oct. 21, 1993

[51] Int. Cl.[6] A61B 8/00

[52] U.S. Cl. 128/660.09; 128/916

[58] Field of Search 128/660.01, 662.02, 128/662.05, 660.09, 916, 653.1

[56] References Cited

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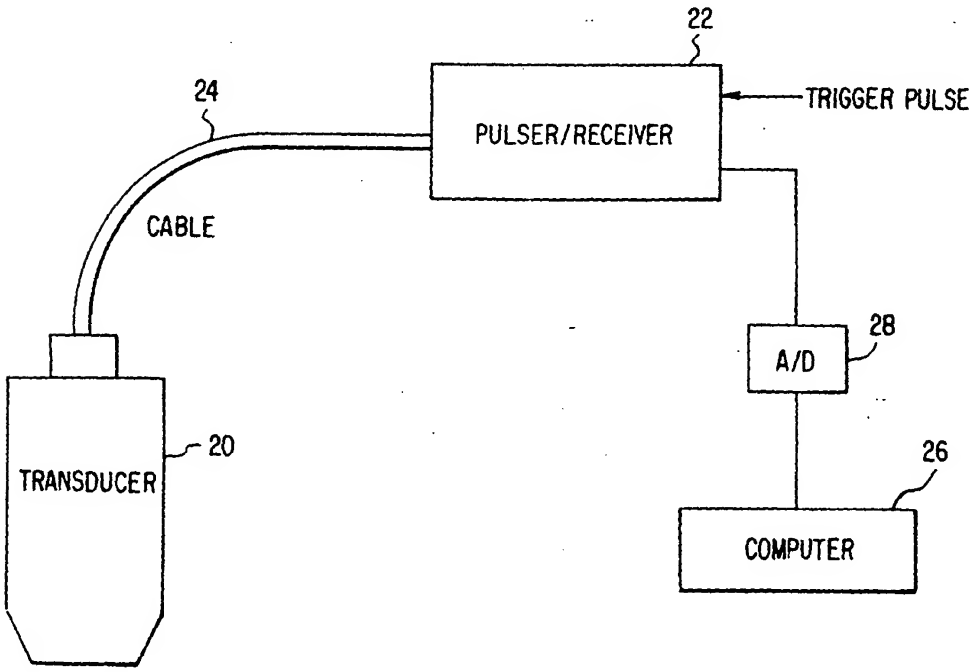
Primary Examiner—George Manuel

Attorney, Agent, or Firm—Kenyon & Kenyon

[57] ABSTRACT

In a coordinate system defined by a coordinate space digitizer, the location of an implanted symmetric object may automatically be determined based on differences between the characteristic acoustic impedances of at least a portion of an implanted object and one or more materials surrounding that object. An A-mode ultrasound transducer may be attached to a coordinate space digitizer or pointing device such that an attached computer tracks the position and orientation of the ultrasound transducer. The reflected radio-frequency (rf) ultrasound signals may automatically be analyzed along with the transducer position and orientation corresponding to each received rf signal to detect the position of the implanted object. The time delay between received ultrasound echoes is used to compute the depth of the object from the transducer. This allows for an accurate determination of the three-dimensional coordinates of the implanted object.

80 Claims, 11 Drawing Sheets



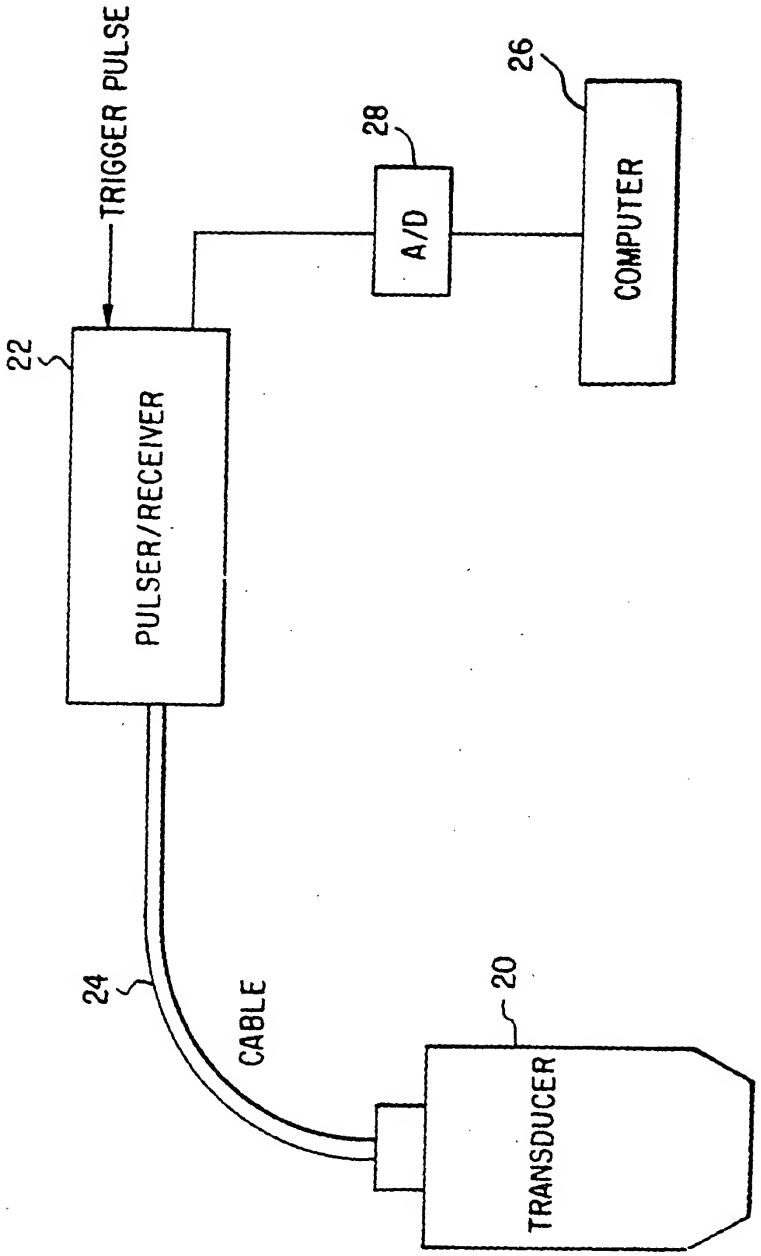

TRIGGER PULSE
26
28
A/D
COMPUTER
22
PULSER/RECEIVER
24
CABLE
20
TRANSDUCER

FIG. 1

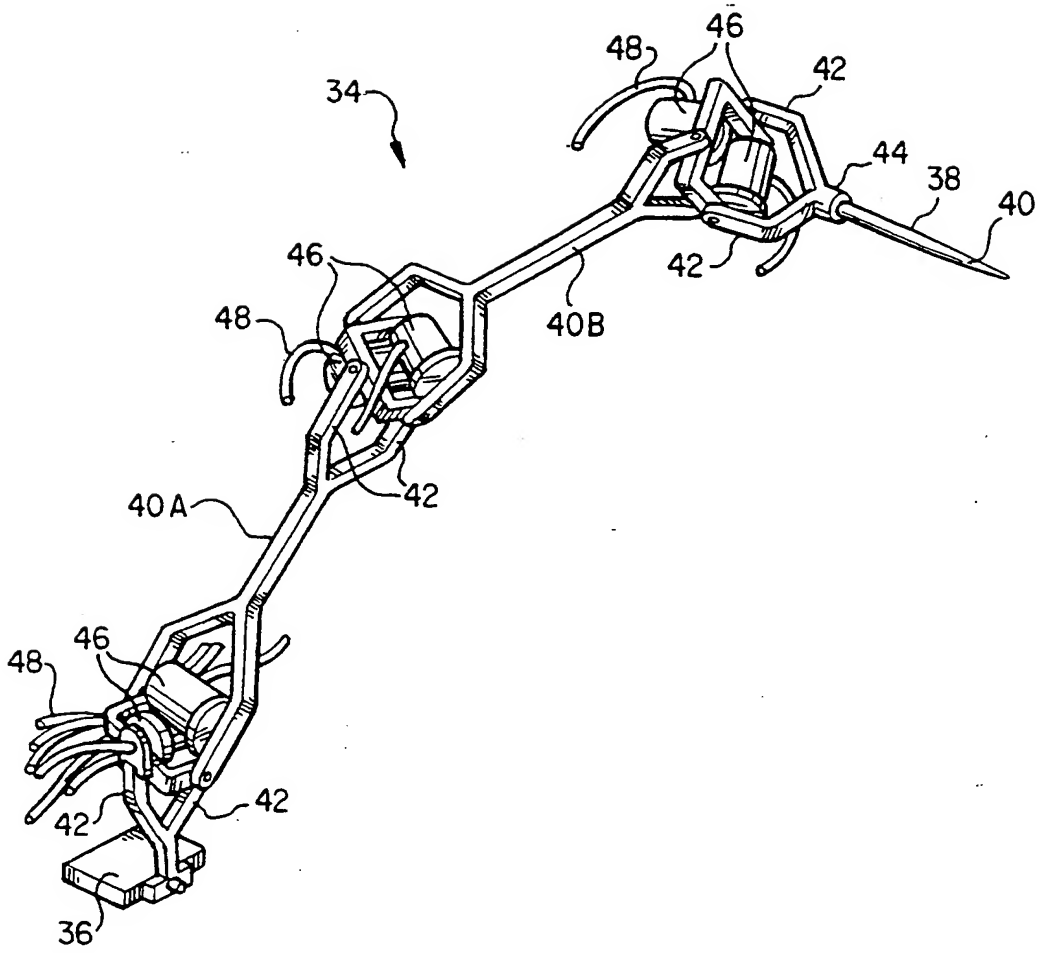


FIG. 2

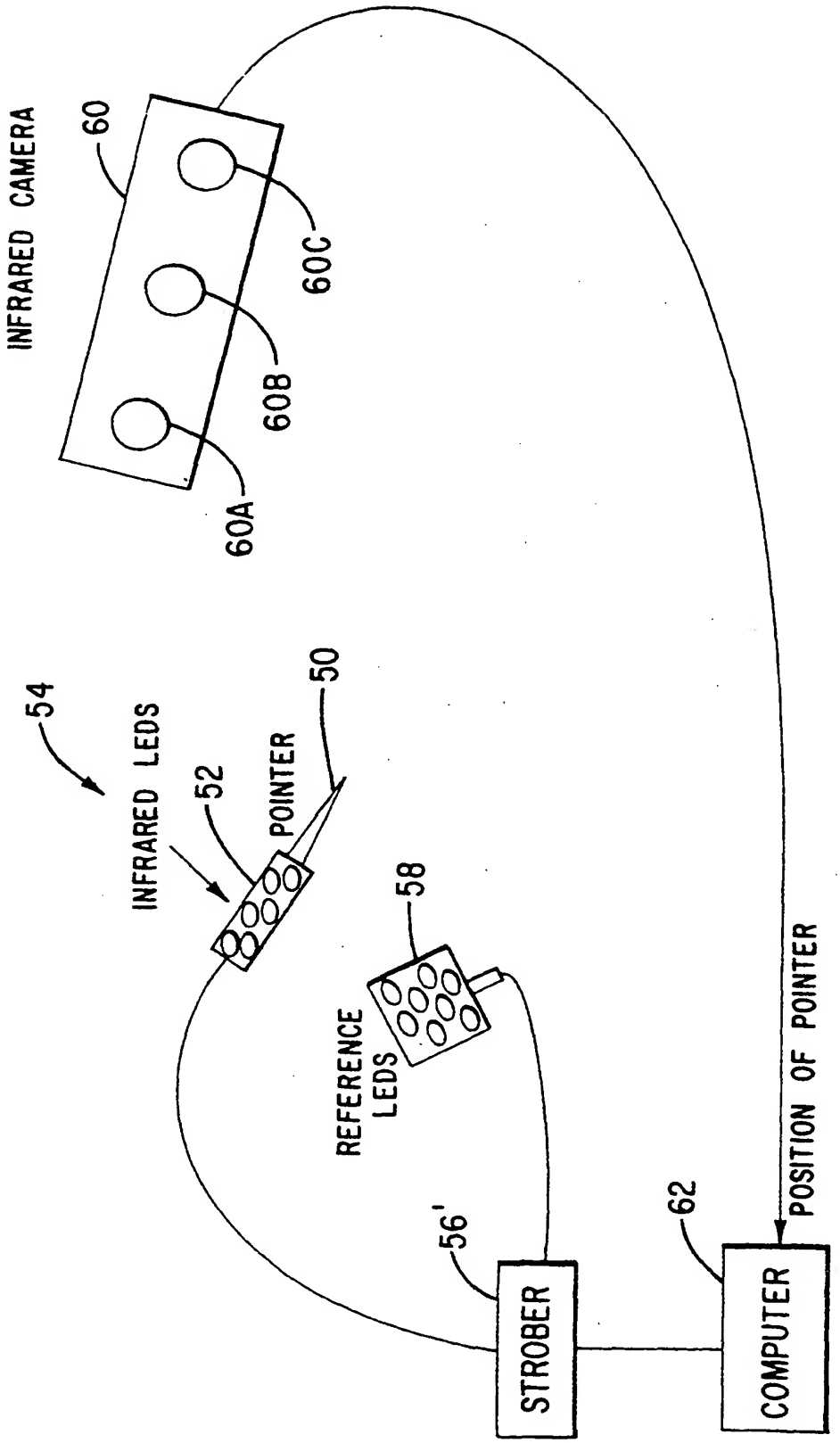

INFRARED CAMERA
60
60C
60B
60A
54
INFRARED LEDS
52
POINTER
50
58
REFERENCE LEDS
56'
STROBER
62
COMPUTER
POSITION OF POINTER

FIG. 3

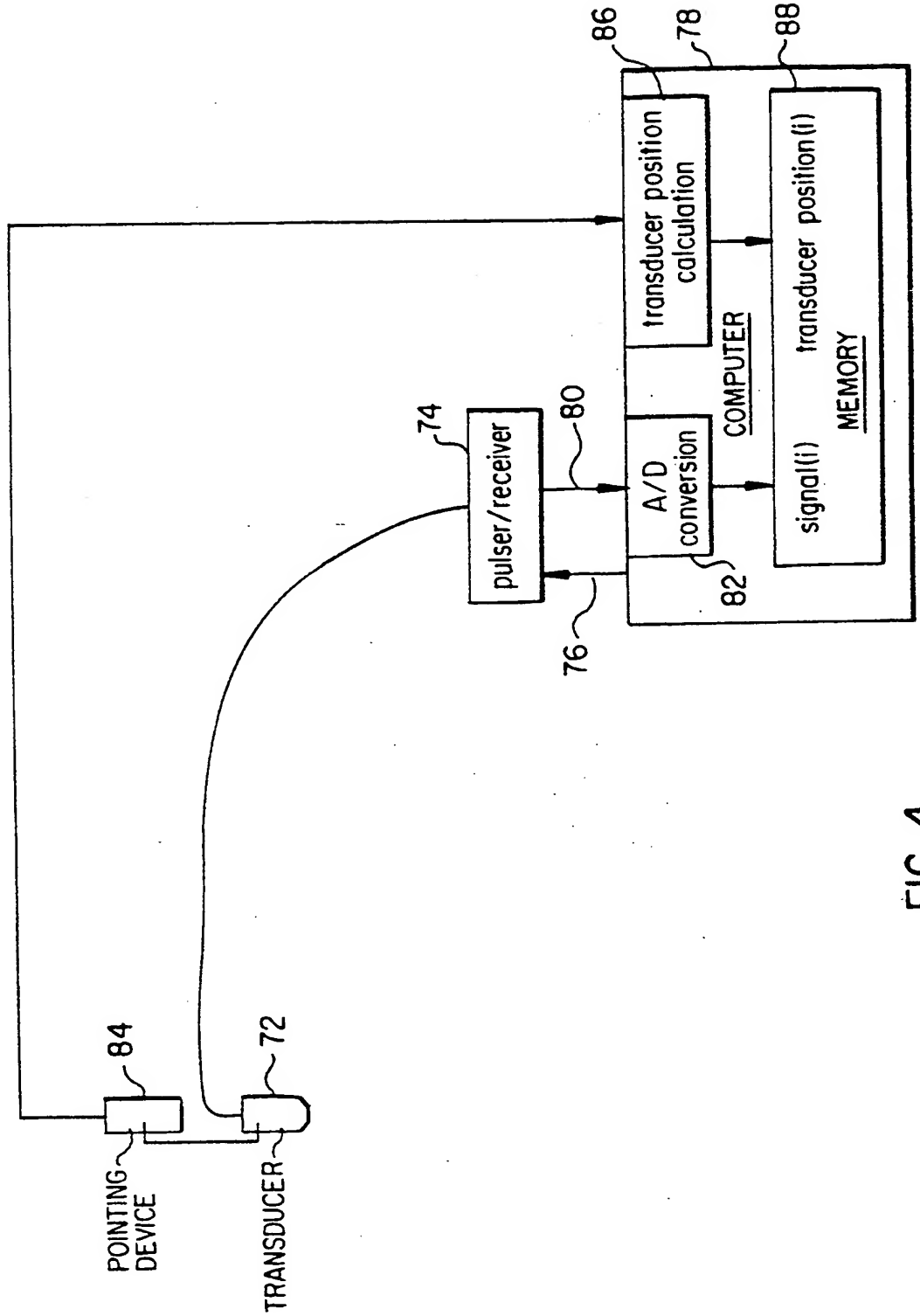

86
78
88
transducer position calculation
transducer position(i)
COMPUTER
MEMORY
signal(i)
74
80
pulser/receiver
A/D conversion
82
76
84
72
POINTING DEVICE
TRANSDUCER

FIG. 4

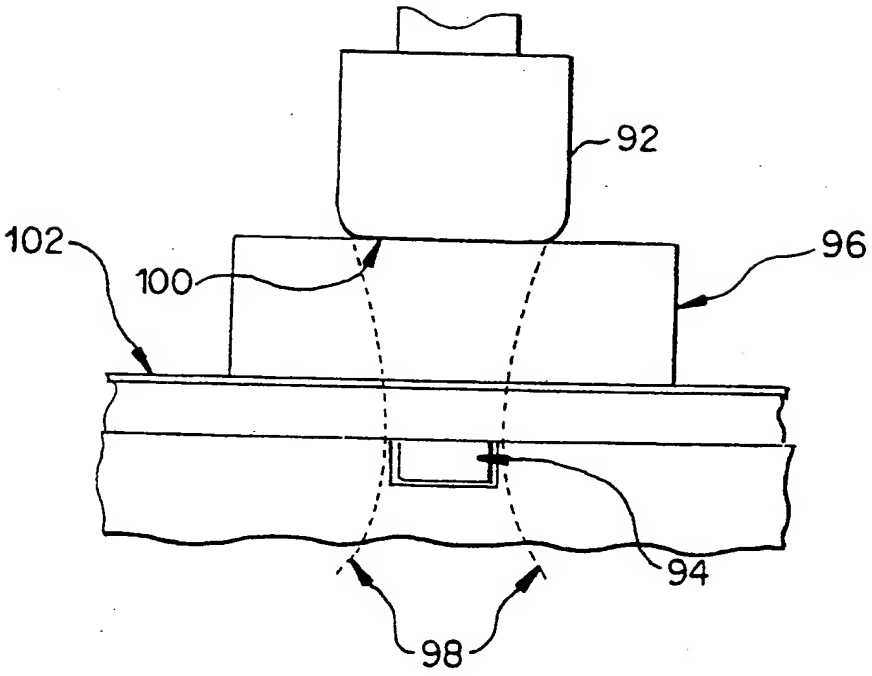


FIG. 5

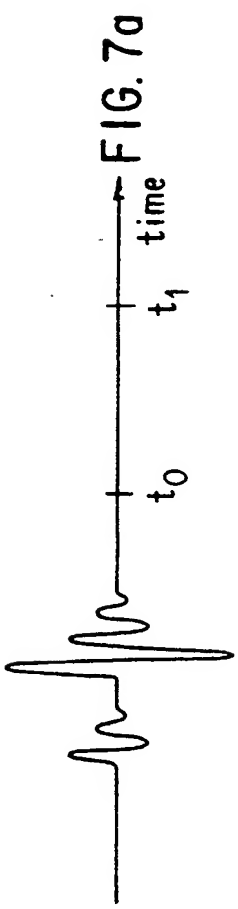


FIG. 7a

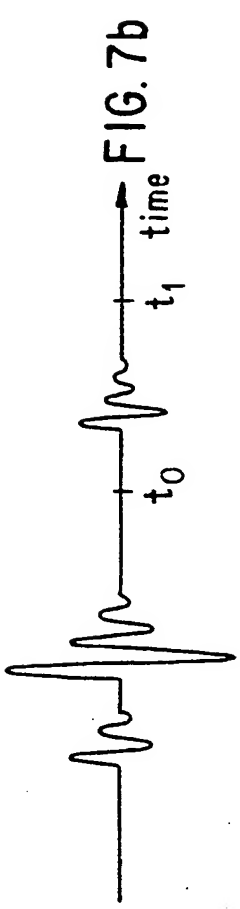


FIG. 7b

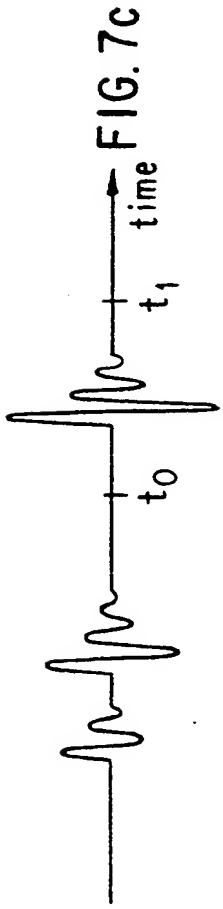


FIG. 7c

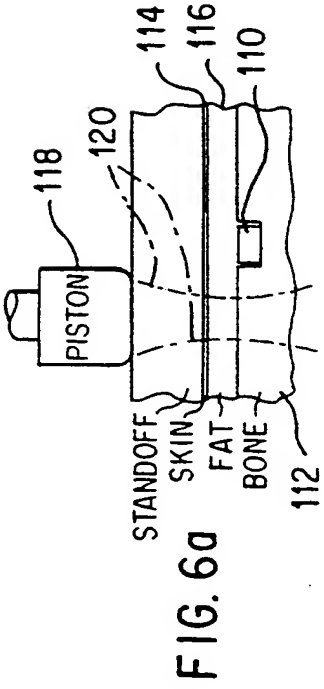


FIG. 6a

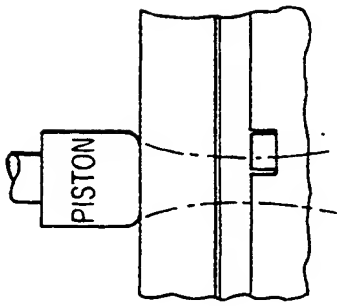


FIG. 6b

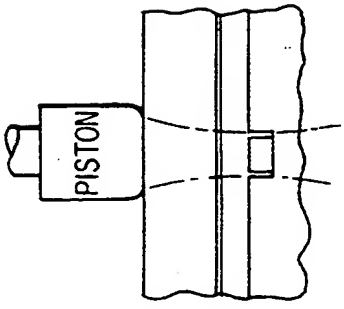


FIG. 6c

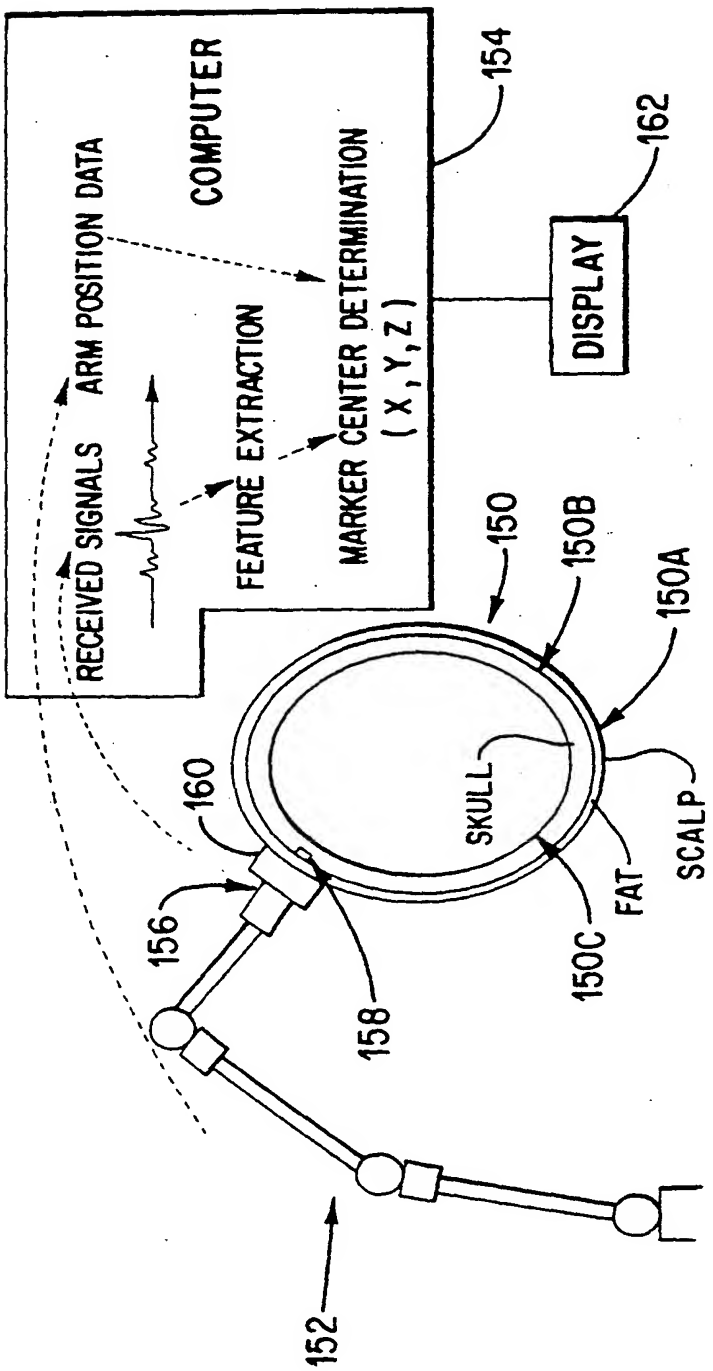

COMPUTER
ARM POSITION DATA
RECEIVED SIGNALS
FEATURE EXTRACTION
MARKER CENTER DETERMINATION
(X , Y , Z)
154
DISPLAY
162
150
150B
150A
160
SKULL
SCALP
FAT
150C
156
158
152

FIG. 12

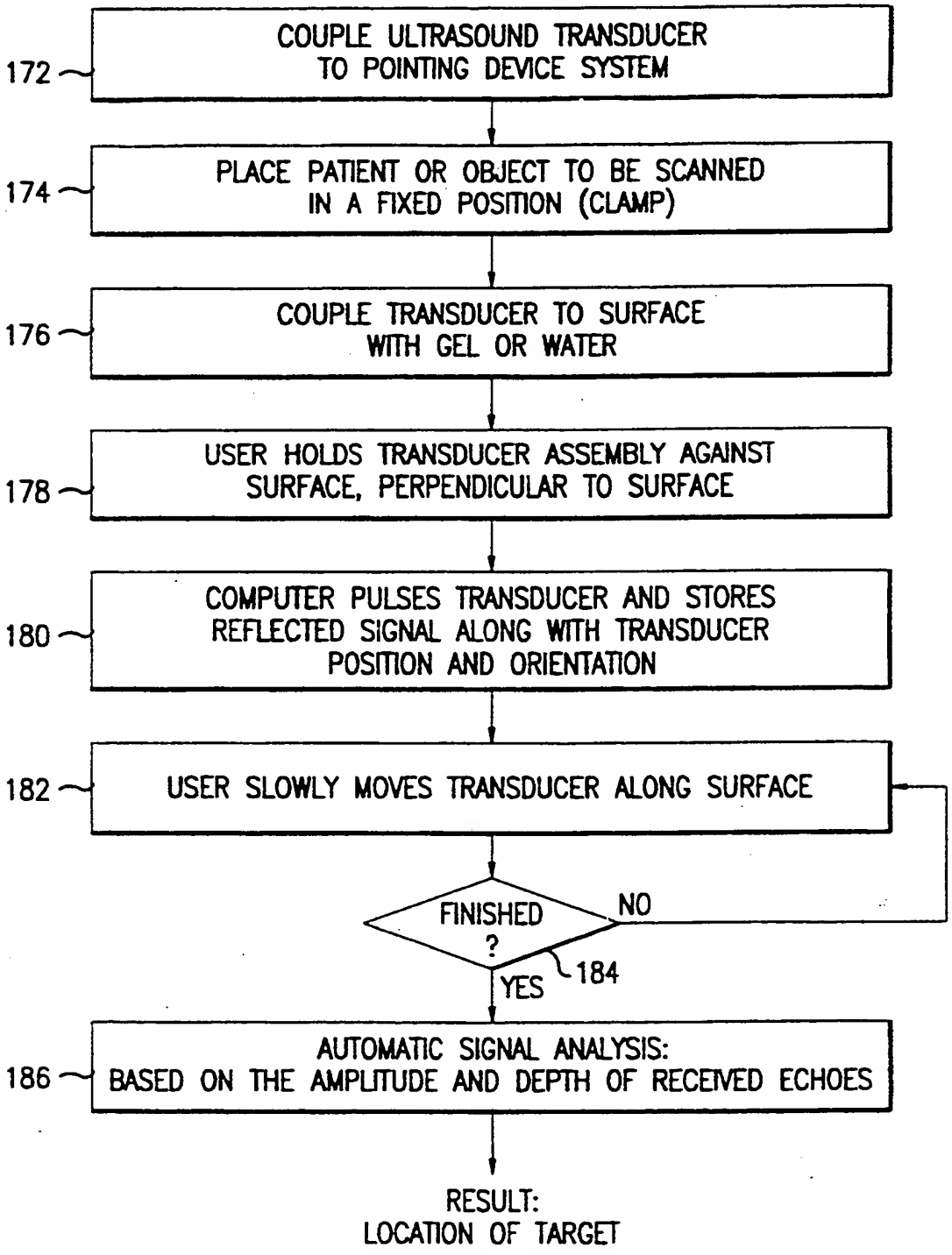

172
COUPLE ULTRASOUND TRANSDUCER TO POINTING DEVICE SYSTEM
174
PLACE PATIENT OR OBJECT TO BE SCANNED IN A FIXED POSITION (CLAMP)
176
COUPLE TRANSDUCER TO SURFACE WITH GEL OR WATER
178
USER HOLDS TRANSDUCER ASSEMBLY AGAINST SURFACE, PERPENDICULAR TO SURFACE
180
COMPUTER PULSES TRANSDUCER AND STORES REFLECTED SIGNAL ALONG WITH TRANSDUCER POSITION AND ORIENTATION
182
USER SLOWLY MOVES TRANSDUCER ALONG SURFACE
FINISHED ?
NO
YES
184
186
AUTOMATIC SIGNAL ANALYSIS: BASED ON THE AMPLITUDE AND DEPTH OF RECEIVED ECHOES
RESULT: LOCATION OF TARGET

FIG.13

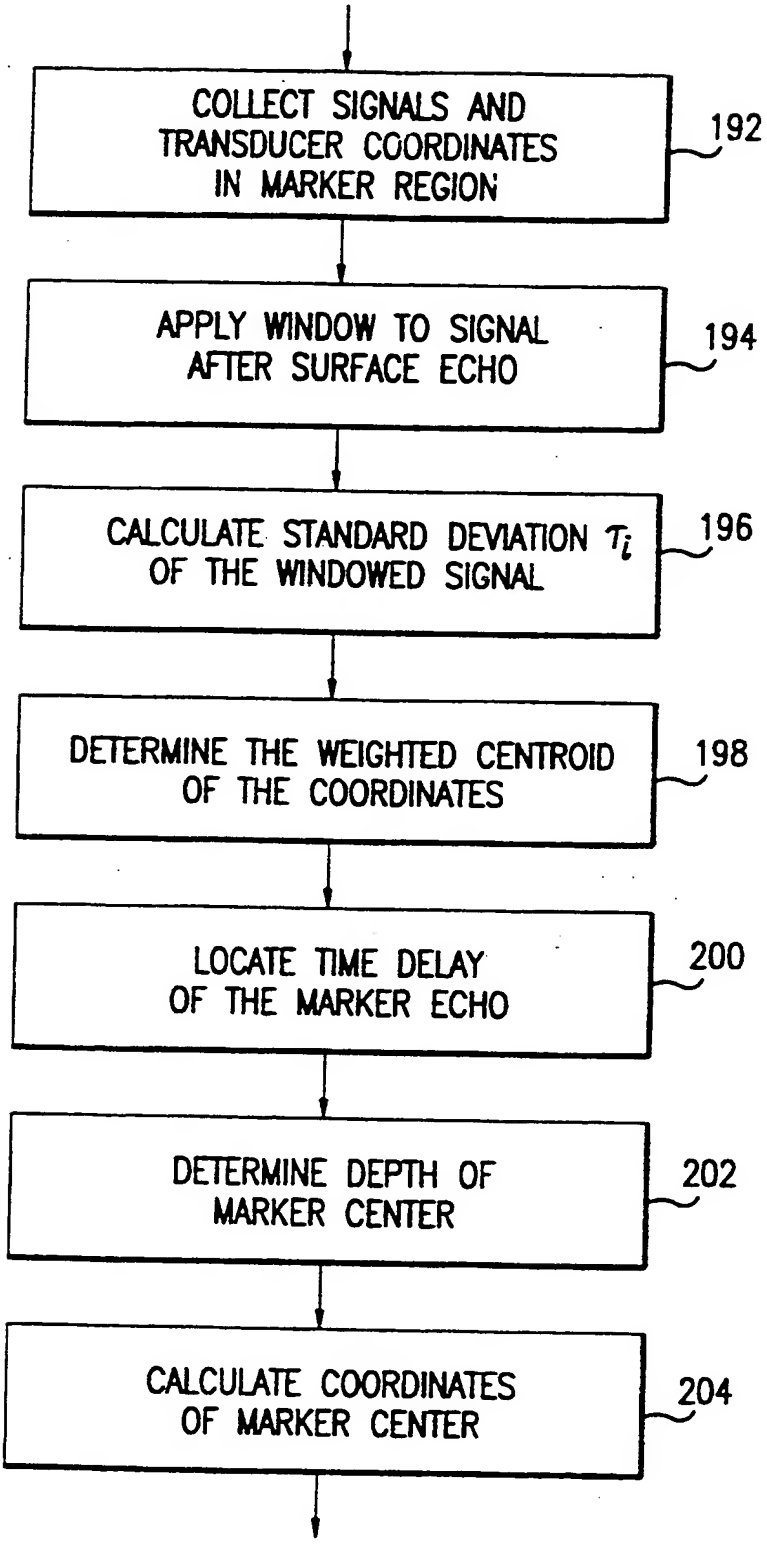

COLLECT SIGNALS AND TRANSDUCER COORDINATES IN MARKER REGION
192
APPLY WINDOW TO SIGNAL AFTER SURFACE ECHO
194
CALCULATE STANDARD DEVIATION τ_i OF THE WINDOWED SIGNAL
196
DETERMINE THE WEIGHTED CENTROID OF THE COORDINATES
198
LOCATE TIME DELAY OF THE MARKER ECHO
200
DETERMINE DEPTH OF MARKER CENTER
202
CALCULATE COORDINATES OF MARKER CENTER
204

FIG.14

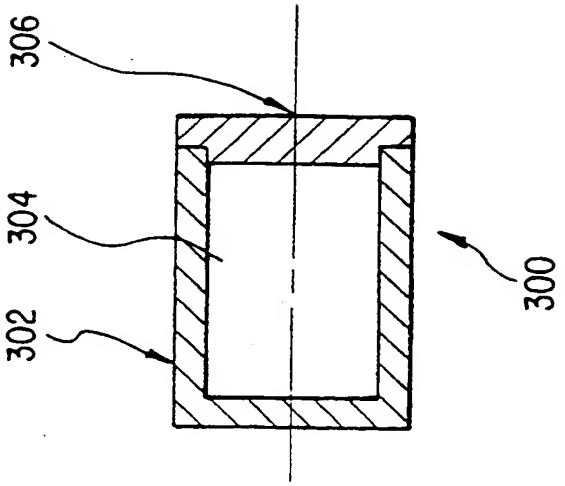


FIG. 15

AUTOMATIC ULTRASONIC LOCALIZATION OF TARGETS IMPLANTED IN A PORTION OF THE ANATOMY

BACKGROUND OF THE INVENTION

The present invention relates to locating a position of an object implanted in a portion of the anatomy, the object having at least a portion which is in a vicinity of a material having a different acoustic impedance than an acoustic impedance of that portion of the object. More particularly, the present invention relates to the localization of implanted targets using amplitude-mode (A-mode) ultrasound techniques and a coordinate space digitizer. Extendable to other applications, the automatic implanted target localization approach may be used to locate small fluid-filled polymer cylinders implanted in human skull, which are preferably flush with the surface of the skull, beneath the scalp and subcutaneous fat. These permanently-implanted cylinders are intended to serve as fiducial markers for the registration of tomographic image volumes with physical space in neurosurgical cases, as well as for tracking a patient over time.

Tomographic imaging modalities, such as computer tomography (CT), magnetic resonance imaging (MRI), and positron emission tomography (PET), produce a three-dimensional image volume as a set of three-dimensional "slices". These image volumes contain the information necessary for surgical or radiotherapeutic planning. Such information may include the three-dimensional distances between sites of interest such as tumors, the location of blood vessels which must be avoided, and lesion margin delineation often superior to that discernible by visual tissue inspection.

However, these three-dimensional volumes are at an arbitrary orientation to each other as well as to the physical patient anatomy. As a result, correlating and comparing the same location in the images is difficult. Also, the surgeon is unable to accurately correlate the detailed image information with the physical anatomy during surgery as a guidance tool.

The correlation of multiple three-dimensional volumes or spaces is referred to as space registration. Space registration establishes a one-to-one mapping between points in the image sets or between points in one or more image sets and physical space. The transformation between coordinate systems is calculated based on the location of a set of at least three common landmarks, or fiducial markers, in each representation. The actual fiducial point is the geometric center of the fiducial marker. The accuracy of the mapping between coordinate systems depends on the accuracy with which the coordinates of the fiducial markers centers are known in each three-dimensional space.

The fiducial markers provide a frame of reference to make image-to-image or image-to-physical space registration possible. A general technique for using fiducial markers to obtain registration of image data across time is set forth in U.S. Pat. No. 4,991,579 to Allen et al., which is incorporated herein by reference.

Fiducial markers for accurate image-to-physical space registration must be rigidly located and must be composed of materials making them visible in the imaging modalities of interest. U.S. patent application Ser. No. 08/017,167 to McCrory et al., which is incorporated herein by reference, describes the design and composition of fiducial markers for neurosurgical image registration and image-physical space registration, as well as a method for localizing the center of the imaged markers in the image volumes. This patent application describes two types of markers including temporary markers anchored in the skull for rigidity but with the image-visible portion protruding above the scalp, and permanent markers implanted into the skull beneath the scalp and subcutaneous fat.

Internal permanent fiducial markers allow the comparison of images over time for follow-up therapy. Permanent markers also allow the delivery of fractionated radiotherapy, in which small doses of radiation are administered in multiple sessions to maximize the dose delivered to the target lesion while minimizing the damage to surrounding tissue. Fractionated radiotherapy requires the same fiducial framework to be present, in an unchanged position, for each treatment so that the radiation beam may be properly directed relative to the fiducial markers as determined from the pre-treatment images. Temporary markers and stereotactic frames can neither remain in position long enough nor be re-affixed accurately to satisfy this requirement.

In addition to a fiducial marking approach, image-to-physical space registration requires a method for establishing the coordinate system in physical space. Several coordinate space digitizers including specific pointing devices have been developed which define a coordinate space and pass the three-dimensional coordinates of the endpoint to an attached computer. As an example, an articulated arm has joints that track the angular position of each link, allowing the attached computer to calculate the endpoint coordinates. A less cumbersome alternative is a wand with infrared light emitting diodes (LEDs) along its shaft in which the LEDs are strobed in sequence and an infrared camera attached to a computer notes the location of the LEDs relative to a set of reference LEDs in a fixed position.

Also required of a system which correlates neurological image space and physical space is a means for locating the center of the fiducial markers in the coordinate system defined by the pointing device. For markers visible outside the scalp, the fiducial position can be recorded simply by touching the fiducial marker with the pointing device. For permanent, subcutaneously-implanted markers, however, locating the precise three-dimensional position of the marker is much more challenging. A target localization method to address this task has previously become necessary in the field of registration of image volumes with physical space in neurosurgical cases.

Once the markers have been located in the preoperative image sets stored on a computer as well as in the physical space defined by a pointing device attached to the same computer, the system display can provide interactive surgical guidance. The location of the endpoint of the pointing device is indicated on a display of the appropriate slice through the image volumes. Such an interactive, image-guided surgery system using an articulated arm as a pointing device is described in U.S. Pat. No. 5,142,930 to Allen et al., which is incorporated herein by reference.

U.S. Pat. No. 5,197,476 to Nowacki et al., which is also incorporated herein by reference, discloses an ultrasound probe coupled with an infrared LED-based pointing device to locate a target in a living body. A three-dimensional frame containing a plurality of infrared lights is placed on a table. A computer strobes the

infrared lights and the position of the infrared lights is monitored by a pair of infrared sensitive cameras and stored in the computer. A hand held ultrasonic probe is provided with a plurality of infrared lights so that the probe can be monitored by the cameras. The computer compares the positions of the probe lights with the initial position of the frame infrared lights to accurately determine the position of the probe so that the position of the target in the body may be displayed on a computer monitor. The approach disclosed in the Nowacki et al. patent employs a brightness-mode (B-mode) ultrasound imaging and requires a trained expert to visually recognize when the desired target is displayed.

SUMMARY OF THE INVENTION

The present invention relates to a method and apparatus for automatically localizing the three-dimensional coordinates of a symmetrically-shaped implanted target. The ultrasonic method is based on differences in the characteristic acoustic impedance (Z_0) of the target and that of the surrounding material or surrounding materials as well as on the time delays between ultrasonic echoes.

The present invention uses A-mode ultrasound which produces one-dimensional, time domain "signals". Echoes received by the transducer are indicated as deflections in the signal and the amplitude of these deflections is proportional to the difference in acoustic impedances of the materials which form the interface from which the echo arose. In contrast, B-mode ultrasound imaging produces a two-dimensional image such as that provided on a video display, where each displayed line of the image corresponds to a single A-mode signal acquired from adjacent positions. The brightness or intensity of each pixel or dot along that line corresponds to the amplitude of received echoes.

According to an embodiment of the present invention, an A-mode ultrasonic transducer is attached to a pointing device which digitizes the coordinates of physical space. The automatic target detection algorithm is implemented on a computer that receives the reflected ultrasound signals along with the transducer position and orientation information. A position of the target or fiducial marker in a coordinate system of the pointing device is determined in response to the received echoes and the position and orientation of the transducer.

The present invention is based on the following ultrasonic principles. When an ultrasonic pulse is emitted from a transducer, it propagates through a material until it reaches an interface with a material of different characteristic acoustic impedance. At this interface, a portion of the power of the pulse is reflected back to be received by the transducer and the remainder propagates on to deeper interfaces. The ratio of reflected to transmitted power is proportional to the square of the difference in the characteristic acoustic impedances of the two materials. The time delay between the echoes received by the transducer may be multiplied by the speed of sound in the intervening material to obtain the distance traveled by the ultrasonic pulse and the speed of sound in the intervening material.

The present invention provides accurate localization, appropriate for localizing the center of small targets such as those a few millimeters in diameter. Accuracy is critical for the described application of the invention to image-to-physical space registration. An example of a fiducial marker design for that application is a cylinder, 3 mm in diameter, 4 mm in height, consisting of a polymer shell and an aqueous solution. Such a marker would be implanted into the skull flush with the surface of the skull, beneath the skin and subcutaneous fat.

BRIEF DESCRIPTION OF THE DRAWINGS

Other features and advantages of the present invention will become apparent from the following description taken in conjunction with the attached drawings.

FIG. 1 illustrates an A-mode ultrasound transducer arranged with associated hardware according to an embodiment of the present invention.

FIG. 2 is a mechanical articulated arm which may be used as a pointing device in an image-guided neurosurgery system.

FIG. 3 is a schematic of the components of an optical pointing device system.

FIG. 4 illustrates the acquisition of signal and position information from an ultrasound transducer coupled to a pointing device according to an embodiment of the present invention.

FIG. 5 illustrates a position of a transducer piston near a target and the use of an ultrasound standoff pad in placing the target in the optimal range of the ultrasound beam.

FIG. 6, which includes FIG. 6(*a*), FIG. 6(*b*), and FIG. 6(*c*), illustrates different relative positions of a transducer and a target during an ultrasound location method according to an embodiment of the present invention.

FIG. 7(*a*), FIG. 7(*b*) and FIG. 7(*c*), which together comprise FIG. 7, illustrate a correspondence between the positions illustrated in FIG. 6(*a*), FIG. 6(*b*) and FIG. 6(*c*) and corresponding received reflections including a target interface echo signal on which the automatic localization technique according to the present invention is based.

FIG. 8, which includes FIG. 8(*a*) and FIG. 8(*b*), illustrates a low-impedance implanted fiducial marker for a neurosurgical space registration application of the present invention.

FIG. 9, which includes FIG. 9(*a*) and FIG. 9(*b*), illustrates expected received signals for the relative positions of the transducer piston and the fiducial marker illustrated in FIG. 8(*a*) and 8(*b*), respectively.

FIG. 10 illustrates an embodiment of the present invention in which a low impedance fiducial marker with a high impedance layer is used.

FIG. 11 illustrates expected received signals for the relative positions of the transducer piston and the fiducial marker illustrated in FIG. 10.

FIG. 12 illustrates an embodiment of the present invention relating to surgical applications thereof.

FIG. 13 illustrates a flow chart of an automatic ultrasonic localization method of implanted targets according to an embodiment of the present invention.

FIG. 14 illustrates one embodiment of an automatic signal analysis determination which may be used in implementing an embodiment of the present invention.

FIG. 15 illustrates a fiducial marker which may be used in implementing an embodiment of the present invention.

DETAILED DESCRIPTION

According to an embodiment of the present invention, an A-mode ultrasound transducer possessing a rotationally-symmetric beam (for example, a circular, single-crystal piston transducer) is attached to a pointing device such that the connected computer may track

the coordinates and orientation of the transducer as the transducer is moved through space.

FIG. 1 illustrates an example of an A-mode ultrasound transducer 20 with a beam (not illustrated) having a circular cross-section. In response to a trigger pulse, a pulser/receiver 22 provides an electrical trigger signal to the transducer 20 via a cable 24. In response to the electrical trigger signal transducer 20 emits an ultrasonic pulse. The transducer 20 then receives reflected acoustic waves and converts them to electrical time-domain signals, passing them to the pulser/receiver 22 via cable 24. These signals may be transferred to a computer 26 via an analog-to-digital (A/D) converter 28. The trigger pulse signal input to pulser/receiver 22 may be provided by computer 26.

According to an embodiment of the present invention, an operator translates the transducer 20 along a surface of interest perpendicular to the surface and with the aid of a coupling medium, such as water, acoustic scanning gel, or mineral oil. The computer 26 pulses the transducer 20 via pulser/receiver 22 and receives the reflected echoes along with the corresponding transducer location and orientation as determined by the pointing device system. As the emitted acoustic waves travel through layers of material with similar characteristic acoustic impedances (Z_0), little power is reflected back to the transducer 20. As a result those received echoes have a small amplitude.

When the ultrasound beam reaches an interface with an object whose acoustic impedance Z_0 is much different, a larger-amplitude echo results. Therefore, if an implanted target is composed of a material with an acoustic impedance Z_0 that is much greater or much less than the layers above or below it, a "target interface echo" will result. If enough prior knowledge exists to ensure that no other similar reflectors can be located in the expected depth range of the imbedded target, then the target can be automatically detected according to the present invention based on the presence of such a target interface echo.

The amplitude of the target interface echo will be a maximum when the largest portion of the ultrasound beam reaches and is reflected from that interface. If the ultrasound beam has a diameter greater than or approximately the same as the diameter of the target, the maximum echo amplitude occurs when the transducer is centered over the target. This theory, coupled with the position of the transducer corresponding to each received signal, may be used to map out the position of the target in the lateral plane, the plane perpendicular to the ultrasound beam. To then determine the depth of the target from the transducer face, the present invention detects the time delay between received echoes. Based on the knowledge of the speed of sound in the intervening materials, the direction of the transducer's orientation may be extrapolated along to estimate the three-dimensional location of the target.

One example of a pointing device which may be used to implement an embodiment of the present invention is an articulated arm employing encoders at each joint which track the angles between the arm's segments. FIG. 2 illustrates an example of such a pointing device system to which an ultrasound transducer may be attached according to an embodiment of the present invention. FIG. 2 illustrates an articulated arm 34 such as that used in the interactive, image-guided neurosurgery system described in U.S. Pat. No. 5,142,930 to Allen et al. As illustrated in FIG. 2, an external arm 34 is fixed to a base 36, which is movably fixed to some location. The arm 34 carries a tool 38 including an end tip 40 which is changeable and, for purposes of the present invention, is a pointing device which may include an attached ultrasound unit. A sensor (not illustrated in FIG. 2) which comprises an ultrasonic detector may be attached in place of the tool 38 and the end tip 40 of the tool 38 in order to practice the present invention. The arm 34 has two arm lengths 40A, 40B. The first arm length 40A is coupled to the base by two gimbal joints 42. The first arm length 40A therefore has two degrees of motion, as provided by the two gimbal joints 42.

A second arm length 40B is coupled to the first arm length 40A by a second pair of gimbal joints 42. The second pair of gimbal joints 42 provides the second arm length 40B with two additional degrees of motion. The second arm length 40B therefore has four degrees of motion relative to the base 36 of the arm 34.

A tool holder 44 is coupled to the second arm length 40B through a pair of gimbal joints 42. The tool holder 44 can hold any of a number of different tools, including a pointer, an ultrasound unit, a surgical laser, a biopsy probe, a radiation beam collimator, etc. In an embodiment of the present invention, the tool held by the tool holder 44 is an ultrasound unit (not illustrated in FIG. 2). A third pair of gimbal joints 42 provides the tool 38 with two additional degrees of motion, so that the tool 38 has 6 degrees of motion relative to the base 36.

The exact positioning of the tool 38 relative to the base 36 may be monitored by optical encoders 46. One optical encoder 46 is assigned to each gimbal joint 42. Each gimbal joint 42 is individually rotated around its pivot and the optical encoder 46 determines the precise amount of rotation of the gimbal joint 42 around its pivot. The information from each of the six optical encoders 46 is provided to a programmable computer (not illustrated in FIG. 2) via wires 48. The programmable computer can therefore precisely track the movement of the tool 38 relative to the base 36 by keeping track of the individual rotations of the gimbal joints 42 around their pivots.

Another example of a pointing device which may be used to implement the present invention uses infrared light emitting diodes (LEDs) along its shaft. The LEDs are strobed in sequence and an infrared camera attached to a computer notes the location of the LEDs relative to a set of reference LEDs in a fixed position. Such a system which digitizes the coordinates of physical space is the Optotrak/3020 (Northern Digital, Inc., Waterloo, Ontario), illustrated in FIG. 3. One implementation of this system is described in U.S. Pat. No. 5,197,476 issued to Nowacki et al.

The optical system illustrated in FIG. 3 tracks the coordinates and orientation of an object 50 which is illustrated as a pointer, but may be replaced by an A-mode ultrasound transducer to implement an embodiment of the present invention. A shaft 52 extends from object 50 with a plurality of infrared light emitting diodes (LEDs) on its surface. A cable connects the probe 54 (consisting of shaft 52 and end-effector 50) to a strober 56. The probe 54 may be easily and freely manipulated by a user. Mounted in a fixed position near the region of interest is a set of reference infrared LEDs 58 referred to as a rigid body. This set of reference LEDs 58 establishes the origin of a three-dimensional coordinate system in physical space.

The infrared LEDs 52 and 58 are strobed in sequence by strober 56 and are visible to a position sensor 60.

Position sensor 60 includes three linear charge coupled device (CCD) cameras 60A, 60B, 60C with cylindrical optics placed in front of the CCDs to compress the field of view into a single line. The output of each CCD 60A, 60B, 60C is captured by a separate processor (not illustrated) which extracts the object's position in that view. A fourth processor (not illustrated) receives the output of each of the three separate individual processors and triangulates the location of an infrared LED and passes it to a computer 62. Since the infrared LEDs on the shaft 52 are pulsed in a known sequence and configuration, and the length of the end-effector 50 is known, the computer 62 is able to calculate the time-distinct location of the end-effector 50 with respect to the rigid body of reference infrared LEDs 58.

The present invention is not limited to embodiments of the present invention using the pointing devices illustrated in FIG. 2 and FIG. 3. In the present invention, an ultrasound transducer may be attached to any pointing device such that the coordinates and orientation of the transducer are continuously passed to the connected computer, as illustrated, for example, in FIG. 4.

In FIG. 4, each time a transducer 72 is pulsed by a pulser/receiver 74 in response to a trigger signal 76 from computer 78, an ultrasound signal 80 of specific time duration is acquired from pulser/receiver 74, digitized by an A/D converter 82 which may be included in computer 78, and stored as a signal (i) in a memory 88 of the computer 78. The rate at which the A/D converter 82 samples the analog ultrasound signal 80 must be sufficient to adequately represent the signal. The sampling frequency should be at least twice the frequency bandwidth of the ultrasound signal. Simultaneously, the position and orientation of the transducer 72 corresponding to the current signal are acquired from a pointing device 84. A transducer position calculator 86 calculates a transducer position (i) in response to the data provided from pointing device 84 which is stored in memory 88 along with the signal (i). In this manner, signal/position pairs are acquired while a region of an object or anatomy are scanned with the transducer assembly.

To gather signal/position pairs from an object or patient once that object or patient is secured in a fixed position, the transducer assembly is held by a user and moved across a surface of interest, keeping the transducer perpendicular to the surface. FIG. 5 illustrates a position of a transducer piston 92 near a target 94. An ultrasound stand-off pad 96 is used in placing the target 94 in an optimal range of an ultrasound beam 98 (illustrated by a dotted line) of the transducer piston 92. Contact between a transducer face 100 and the surface 102 is maintained with the aid of a coupling medium, such as acoustic gel or water. The ultrasonic stand-off pad 96 may be used to place the expected target depth in an optimal range of the ultrasound beam 98 depending on the expected depth of the implanted target 94 and the type of transducer used. Stand-off 96 may be a layer or "pad" of a gelatinous material or a water path.

In the present invention as described above in reference to FIG.4, as the signal/position pairs are acquired, the signals are automatically analyzed. The signals are analyzed to detect echoes arising from interfaces between materials of different characteristic acoustic impedance (Z_0) as described above. FIG. 6, which includes FIG. 6(*a*), FIG. 6(*b*) and FIG. 6(*c*), illustrates this concept as applied to a low-Z_0 target 110 (e.g., a cylindrical fiducial marker) implanted in a high-Z_0 material 112 (e.g., the human skull) covered by two layers 114, 116 of low-Z_0 material (e.g., scalp and fat). FIG. 6(*a*), FIG. 6(*b*) and FIG. 6(*c*) illustrate the relative position of the transducer piston 118 and the target 110. FIG. 7(*a*), FIG. 7(*b*) and FIG. 7(*c*) illustrate corresponding received reflections of the ultrasound signals for the different relative positions of the transducer piston 118 and target 110 of FIG. 6(*a*), FIG. 6(*b*) and FIG. 6(*c*), respectively. As illustrated, for this case the target interface echo is the echo arising from the interface between the distal surface of the target and the high-Z_0 divot in which the target sits. In FIG. 7(*a*), FIG. 7(*b*) and FIG. 7(*c*), the time interval (from t_0 to t_1) corresponds to the depth of the target interface echo. This time interval is the time interval in which the target interface echo is expected in this example. The reflection signal illustrated in FIG. 7(*c*) is the signal used to determine the depth of the target from the transducer since it corresponds to the accurate relative positioning of the transducer piston 118 and target 110 illustrated in FIG. 6(*c*). The computer 26, 54, or 78, for example, extracts the portion of each signal between t_0 and t_1 and the result is referred to as the windowed signal.

Since the amplitude of the extracted windowed signal is proportional to the proximity of the transducer beam 120 to the center of the target 110, a measure of each windowed signal's amplitude is calculated. For signals whose measured amplitude exceeds a given threshold, the transducer coordinates corresponding to that signal are weighted with that measured amplitude. A centroid is then calculated which estimates the coordinates of the transducer face when it was centered over the target. The signal or signals acquired nearest that centroid are examined and the time index of the leading edge of the target interface echo is detected. Based on the knowledge of the speed of sound in the materials and the geometry of the target, the depth of the target from the transducer face is determined. By adding this depth to the transducer position corresponding to that signal along the orientation of the transducer (also stored along with the signal), the estimated three-dimensional coordinates of the target may be calculated.

As an example, the present invention may be used in image-to-physical space registration for neurosurgical guidance. Once the fiducial markers are implanted in the patient, the preoperative image sets may be acquired. These fiducial markers may be between 1 mm and 10 mm in diameter and are composed of biocompatible materials which are visible in the imaging modalities of interest. The markers are implanted into the skull, flush with the surface of the skull.

Possible marker scenarios include in FIG. 8(*a*) and FIG. 8(*b*) a marker 130 in which small cylinders with a thin shell of low-Z_0 polymer material filled with a water-based solution, or in FIG. 10 a marker 140 which is identical to marker 130 except that a high-Z_0 layer 142 such as a glass marker or "cap" is included on the end of the cylinder which is to be implanted first (i.e., the end of the marker distal to the surface of the skull).

FIG. 9(*a*) and FIG. 9(*b*) illustrate expected received echo signals in response to the respective positions of the transducer piston and fiducial marker 130 illustrated in FIG. 8(*a*) and FIG. 8(*b*), respectively. The received signal in FIG. 9(*b*) includes a marker/bone divot echo portion which is not included in the received signal illustrated in FIG. 9(*a*). This marker/bone divot echo portion is the portion of the signal which is extracted as the windowed signal as discussed above.

As mentioned above, FIG. **10** illustrates a fiducial marker **140** identical to marker **130** except that a high-Z_0 layer **142** such as a glass marker "cap" is included on a distal surface of marker **140**. FIG. **11** illustrates the expected received signal in response to the transducer piston and fiducial marker **140** arrangement of FIG. **10**. The signal illustrated in FIG. **11** includes a marker contents/marker "cap" echo portion which is the portion of the signal extracted as the windowed signal.

In the method of the present invention, each image volume is analyzed to determine the location of the center of each fiducial marker within the three-dimensional coordinate system of the image volume (three or more markers are preferably used in order to more accurately determine coordinate positions). The image volumes and the fiducial locations in each volume are stored on a computer.

In the operating room, the patient **150** illustrated in FIG. **12** is fixed in position for surgery, usually with a head clamp. A pointing device such as interactive, image-guided (IIG) arm **152** is connected to a computer **154** holding the image sets and image fiducial locations. An A-mode ultrasound transducer **156** is attached to the pointing device **152**. A fiducial marker **158** is implanted below the scalp **150A** and subcutaneous fat **150B** into the skull **150C** of the patient **150**. An example of a transducer **156** which may be used in this arrangement is a 10 MHz short-focus piston with a 3 mm-diameter crystal. The pointing device **152** is set up such that the computer **154** continuously calculates the three-dimensional coordinates of the center of the transducer face and the orientation of the transducer. The computer **154** receives the signals from transducer **156** and performs a feature extraction on the received signals in which, for example, the marker/bone divot echo portion or marker contents/marker "cap" echo portion of the signal is extracted. Then computer **154** determines a center position of the marker **158** in response to arm position data received from the pointing device **152** and in response to the feature extraction signal.

In order to implement the present invention, an operator holds the transducer assembly **156** and places it on the scalp **150A** of the patient **150** while maintaining contact between the transducer and scalp with acoustic gel and an ultrasonic standoff **160**. The operator starts a program within computer **154** that pulses the transducer **156** and receives the reflected signals along with the transducer position information. The operator slowly moves the transducer **156** while signal/position information is collected. The received signals are analyzed to isolate the transducer positions for which the ultrasound beam passes through the marker **158**. For space registration purposes, it is the coordinates of the center of the marker **158** which are of interest. As illustrated in FIG. **12**, an estimate of the center of the marker **158** may be determined using a centroid/extrapolation process and stored in the computer **154**.

This process is repeated until all of the markers (preferably three or more) have been localized. To confirm accuracy of the marker center estimates, the distances between the marker centers are compared to the distances between marker centers as determined from, for example, preoperative images. If the error is within an acceptable range, a transformation is calculated mapping the surgical space coordinates of the pointing device into equivalent locations in the image sets.

The transducer **156** is removed from the pointing device **152**, and associated hardware and software are updated via computer **154** to reflect the resulting change in length of the pointing device **152**. The surgeon may then point at locations of interest on the patient and look at a display **162** of the computer **154** to see the corresponding location indicated on the preoperative image sets. In this way, accurate, interactive surgical guidance is provided to the surgeon.

FIG. **13** illustrates a flowchart of an ultrasound method of locating implanted targets according to an embodiment of the present invention. In step **172** the ultrasound transducer is coupled to the pointing device system. The patient or object to be scanned is placed in a fixed position, for example, by clamping the patient or object to a table (step **174**). In step **176**, the transducer is coupled to the surface (for example, the scalp of the patient) with gel or water. The user then holds the transducer assembly against the surface in a manner perpendicular to the surface (step **178**). The computer is used to pulse the transducer and store the received reflected signal along with a transducer position and orientation in step **180**. In step **182**, the user slowly moves the transducer along the surface. A determination is made in step **184** as to whether the user is finished. If the user is not finished, the computer again pulses the transducer and stores the reflected signals in step **180**. Once all pulsing and storing of signal information has been finished as determined in step **184**, the computer provides an automatic signal analysis in step **186** which is based on the amplitude and depth of received echo signals. As a result, the center position of the target may be accurately determined.

FIG. **14** illustrates one embodiment in which the automatic signal analysis determination of step **186** in FIG. **13** may be implemented. It is noted that other methods of automatic signal analysis may be implemented within the scope and spirit of the present invention and FIG. **14** merely illustrates an embodiment thereof.

FIG. **14** includes a step **192** of collecting signals and transducer coordinates in the marker region, including the reflected ultrasound signals and the ultrasound transducer position data. In order to perform this step, the reflected signals are analyzed to detect echoes arising from interfaces between materials of different acoustic impedance. Such reflected ultrasound signals include, for example, those signals illustrated in FIG. **7**(*a*), FIG. **7**(*b*), FIG. **7**(*c*), FIG. **9**(*a*), FIG. **9**(*b*) and FIG. **11**. Step **194** relates to applying a window to the collected signal after a surface echo occurs therein. This window corresponds to the portion between t_0 and t_1 in FIG. **7**(*a*), FIG. **7**(*b*) and FIG. **7**(*c*). The windowed portion of the signal is extracted from the reflected signal. The amplitude of the extracted windowed signal corresponds to the proximity of the transducer beam to the center of the target. The standard deviation σ_i of the windowed signal is calculated in step **196** as an indication of the proximity of the transducer to the center of the target. The weighted centroid of the coordinates of the signal are determined in step **198**, for example, according to the following formulas.

$$x_c = \frac{\sum_i x_i(\sigma_i - \text{threshold})}{\sum_i (\sigma_i - \text{threshold})}$$

$$y_c = \frac{\sum_i y_i(\sigma_i - \text{threshold})}{\sum_i (\sigma_i - \text{threshold})}$$

-continued

$$z_c = \frac{\sum_i z_i (\sigma_i - \text{threshold})}{\sum_i (\sigma_i - \text{threshold})}$$

in which, x_i, y_i and z_i correspond to the ultrasound transducer position data received from the pointing device, signal σ_i relates to the standard deviation calculated in step **196**. The "threshold" value is calculated based on the maximum standard deviation σ_i of a signal which did not pass through the marker. By subtracting this threshold value from each standard deviation value σ_i, the effect on the centroid calculation of collecting signal/position data outside the marker region is minimized. The coordinates x_c, y_c and z_c are the coordinates of the transducer face when it was centered over the target. Together, steps **194**, **196** and **198** perform the localization used to determine the lateral position of the center of the marker.

In step **200**, a time delay of the marker echo is located nearest the position of the centroid coordinates (x_c, Y_c, z_c) to detect the leading edge of the target interface echo. Step **202** determines the depth of the marker center, for example, according to the formula:

$$d_c = d_d - 0.5 * m_h$$

where d_c is the depth from the marker center to the face of the transducer, d_d is the distance from the transducer face to the target interface (e.g., the interface between the marker and the divot in which the marker sits), and m_h is the marker height. The depth of the target from the transducer face may be determined based on the speed of sound in the materials and the geometry of the target. By adding this depth to the transducer position corresponding to that signal along the orientation of the transducer (also stored along with the signal), the estimated three-dimensional coordinates of the target may be calculated. Step **204** calculates the three-dimensional coordinates for the marker center based on the depth of the marker, the determined transducer position, and the ultrasound transducer position data. Steps **200**, **202** and **204** together perform depth localization of the marker for determining the depth of the marker from the position of the transducer.

While the signal analysis determination steps illustrated in FIG. **14** have been described as set forth above, it is pointed out that the present invention is not specifically limited to this marker localization method. The present invention could be practiced by locating the position of the target or fiducial marker using another method which accurately determines the position of the marker.

FIG. **15** illustrates a fiducial marker **300** which may be used in implementing the present invention. The fiducial marker **300** may be left implanted, for example, entirely beneath the skin for extended periods of time. The marker **300** comprises a cylinder **302** defining a space **304** into which may be placed, for example, one or more imaging agents. A cylindrical shape is preferred for marker **300**, because this shape minimizes the size of the incision that must be made for the marker's insertion. It is also the shape that best corresponds to the hole that may be drilled in a bone to accommodate the marker. In any case, it is preferred that the marker at least be symmetrical in shape. The body of the cylinder is sealed off with a cap **306** or is otherwise sealed. The body is preferably constructed of an organic polymer known to be well tolerated by the body for extended periods of time, such as polymethyl methacrylate (PMMA), high density polyethylene, or ceramics such as zirconium oxide and aluminum oxide. The entire marker assembly is small enough for long-term implantation into bone without causing distortion of the bone over time. One exemplary size provides for the marker to be 4 mm in diameter and 3 mm in height. The marker may be implanted in a human skull, flush with the surface of the skull, beneath the fat and subcutaneous fat. Additionally, the cap **306** of the marker may include a high acoustic impedance layer such as a glass marker or "cap" similar to the marker illustrated in FIG. **10**.

The cylindrical marker defined above is only one example of possible target composition, geometry and implantation amenable to localization using the present invention. Possible embodiments of implanted targets which may be used to implement the present invention include the following: A low-Z_0 target implanted under one or more low-Z_0 layers but into a high-Z_0 layer, flush with the surface of that layer. The target echo in that case is the interface between the target and the high-Z_0 divot in which the target sits. Additional possibilities include a low-Z_0 target implanted in high Z_0 materials, and a high-Z_0 target implanted in a low-Z_0 material.

The present invention may be embodied in other specific forms other than that specifically disclosed in this application without departing from its spirit or essential characteristics. For example, while a specific fiducial marker has been described in reference to FIG. **15**, the present invention is not limited to localization of this particularly described implanted marker and may be implemented to detect, for example, any implanted target.

What is claimed is:

1. A method for automatically locating a position of a first object in a physical space, comprising steps of:

attaching a transducer to a coordinate space digitizer having a defined coordinate system in said physical space;

pulsing said transducer while translating it along a surface of interest near said first object;

receiving echoes from said pulsed transducer arising from a difference in an acoustic impedance of a first portion of the first object and an acoustic impedance of a material located near said first portion of the first object; and

determining automatically a coordinate position in said defined coordinate system of said first object in response to said received echoes and a position and orientation of said transducer.

2. A method according to claim **1**, wherein said first object is a fiducial marker.

3. A method according to claim **2**, wherein said fiducial marker is implanted in a bone.

4. A method according to claim **3**, wherein said bone is a skull of a patient and said surface of interest is a scalp of said patient.

5. A method according to claim **1**, wherein said material located near said first portion of the first object is a second object having an acoustic impedance different from the acoustic impedance of the first portion of the first object.

6. A method according to claim **1**, wherein said material located near said first portion of the first object is a second portion of the first object having an acoustic

impedance different from the acoustic impedance of the first portion of the first object.

7. A method according to claim 1, wherein said pulsing and receiving steps are repeated for a plurality of positions and orientations of said transducer and said determining step determines a position of said first object in response to said plurality of receiving steps and said plurality of positions and orientations of said transducer.

8. A method according to claim 1, wherein said determining step further comprises steps of:

determining automatically a time delay between said received echoes; and

determining automatically a physical depth of said first object from said transducer.

9. A method according to claim 1, wherein said method determines the coordinate position of the first object in said physical space so that each location in physical space is mapped into a corresponding location in an image volume of said physical space.

10. A method according to claim 9, wherein said physical space to image volume mapping is providing for interactive, image-guide surgery or fractionated radiotherapy.

11. A method according to claim 1, wherein said transducer is an ultrasound transducer.

12. A method according to claim 1, wherein said determining step determines a position of said first object in response to said received echoes, a correspondence between a time delay between received echoes and a physical depth between said first object and said transducer, and a position and orientation of said transducer.

13. A method according to claim 1, further comprising a step of determining a position of said first object in response to said received echoes and a corresponding position and orientation of said transducer which corresponds to a position of said transducer during said pulsing step.

14. A method according to claim 1, wherein said determining step comprises steps of:

extracting a windowed surface echo signal from said received echoes;

calculating a standard deviation of said extracted windowed surface echo signal;

calculating a weighted centroid of positions of said transducer;

locating a time delay of said windowed surface echo signal nearest the weighted centroid;

calculating a depth of a center of the first object in response to a depth of an interface in which said first object is located and a height of said first object; and

calculating coordinates in said coordinate system of the center of the first object in response to said calculated depth and said time delay.

15. A method according to claim 1, wherein said receiving step receives echoes from said pulsed transducer arising from differences in acoustic impedances of said first portion and different portions within said first object.

16. A method according to claim 1, wherein said acoustic impedance of said first portion of said object is lower than said acoustic impedance of said material located near said first portion of said first object.

17. A method according to claim 1, wherein said first object is a symmetric object.

18. A method according to claim 17, wherein said first object is cylindrical in shape.

19. A method for automatic amplitude-mode ultrasound location of an implanted fiducial marker in a physical space, comprising steps of:

attaching an ultrasound transducer to a coordinate space digitizer having a defined coordinate system in said physical space.;

pulsing said ultrasound transducer while translating it along a surface of interest near said implanted fiducial marker;

receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a first portion of the implanted fiducial marker and an acoustic impedance of a material located near said first portion of the implanted fiducial marker; and

determining automatically a coordinate position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer.

20. A method according to claim 19, wherein said fiducial marker is implanted in a bone.

21. A method according to claim 20, wherein said bone is a skull of a patient and said surface of interest is a scalp of said patient.

22. A method according to claim 19, wherein said material located near said first portion of the implanted fiducial marker is a material in which said implanted fiducial marker is implanted.

23. A method according to claim 19, wherein said material located near said first portion of the implanted fiducial marker is an object separate from said implanted fiducial marker having an acoustic impedance different from the acoustic impedance of the first portion of the implanted fiducial marker.

24. A method according to claim 19, wherein said material located near said first portion of the implanted fiducial marker is a second portion of the implanted fiducial marker having an acoustic impedance different from the acoustic impedance of the first portion of the implanted fiducial marker.

25. A method according to claim 24, wherein said implanted fiducial marker comprises a marker body enclosing a space, and wherein the second portion of the implanted fiducial marker is a portion of the marker body of said implanted fiducial marker.

26. A method according to claim 25, wherein the acoustic impedance of the portion of the marker body is higher than the acoustic impedance of the first portion of the implanted fiducial marker.

27. A method according to claim 25, wherein said enclosed space contains one or more imaging agents visible in at least one imaging modality.

28. A method according to claim 25, wherein said portion of the marker body is a cap of the marker body, said marker body and cap enclosing said space.

29. A method according to claim 25, wherein said marker body is cylindrical and said portion of the marker body is one end of the cylindrical marker body.

30. A method according to claim 19, wherein said pulsing and receiving steps are repeated for a plurality of positions and orientations of said transducer and said determining step determines a position of said implanted fiducial marker in response to said plurality of receiving steps and said plurality of positions and orientations of said transducer.

31. A method according to claim 19, wherein said determining step further comprises steps of:

determining a time delay between said received echoes; and

determining a physical depth of said implanted fiducial marker from said transducer.

32. A method according to claim 19, wherein said method determines the coordinate position of the first object in physical space so that each location in physical space may be mapped into a corresponding location in an image volume of said physical space.

33. A method according to claim 29, wherein said physical space to image volume mapping is provided for interactive, image-guided surgery or fractionated radiotherapy.

34. A method according to claim 19, wherein said determining step determines a position of said implanted fiducial marker in response to said received echoes, a correspondence between a time delay between received echoes and a physical depth between said implanted fiducial marker and said transducer, and a position and orientation of said ultrasound transducer.

35. A method according to claim 19, further comprising a step of determining a position of said implanted fiducial marker in response to said received echoes and a corresponding position and orientation of said transducer which corresponds to a position of said ultrasound transducer during said pulsing step.

36. A method according to claim 19, wherein said determining step comprises steps of:

extracting a windowed surface echo signal from said received echoes;

calculating a standard deviation of said extracted windowed surface echo signal;

calculating a weighted centroid of positions of said ultrasound transducer;

locating a time delay of said windowed surface echo signal nearest the weighted centroid;

calculating a depth of a center of the implanted fiducial marker in response to a depth of an interface in which said implanted fiducial marker is located and a height of said implanted fiducial marker; and

calculating coordinates in said coordinate system of the center of the implanted fiducial marker in response to said calculated depth and said time delay.

37. A method according to claim 19, wherein said receiving step receives echoes from said pulsed ultrasound transducer arising from differences in acoustic impedances of said first portion and different portions within said implanted fiducial marker.

38. A method according to claim 19, wherein said acoustic impedance of said first portion of said implanted fiducial marker is lower than said acoustic impedance of said material located near said implanted fiducial marker.

39. A method according to claim 19, wherein said implanted fiducial marker is a symmetric object.

40. A method according to claim 39, wherein said implanted fiducial marker is cylindrical in shape.

41. A method according to claim 19, wherein said coordinate space digitizer includes a pointing device.

42. A method according to claim 19, wherein said method automatically locates coordinate positions of a plurality of implanted fiducial markers.

43. A method according to claim 42, wherein the method determines the coordinate positions of the plurality of the fiducial markers in physical space, said method further comprising a step of mapping each location in physical space into a corresponding location in an image volume of said physical space in response to the determined coordinate positions of the plurality of fiducial markers.

44. A system for locating a position of a first object in a physical space, comprising:

a coordinate space digitizer having a defined coordinate system in said physical space;

an ultrasound transducer connected with said coordinate space digitizer;

a pulser pulsing said ultrasound transducer while translating it along a surface of interest located near said first object;

a receiver receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a first portion of the first object and an acoustic impedance of a material located near said first portion of the first object; and

means for automatically determining a coordinate position in said defined coordinate system of said first object in response to said received echoes and a position and orientation of said ultrasound transducer.

45. A system according to claim 44, wherein said first object is an implanted fiducial marker.

46. A system according to claim 45, wherein said material is a second object in which said implanted fiducial marker is implanted.

47. A system according to claim 45, said implanted fiducial marker including a housing containing a cavity, said cavity containing said first portion, wherein a second portion of the implanted fiducial marker includes said material located near said first portion.

48. A system according to claim 47, wherein said second portion is one end of the housing of the implanted fiducial marker.

49. A system according to claim 48, wherein said one end of the housing comprises a cap of the housing, said housing and cap enclosing said cavity.

50. A system according to claim 47, wherein said cavity contains one or more imaging agents visible in at least one imaging modality.

51. A fiducial marker assembly comprising an imaging marker assembly including a housing containing a cavity, said imaging marker assembly comprising:

a first portion including a first material having a first acoustic impedance; and

a second portion including a second material having a second acoustic impedance which is different than said first acoustic impedance.

52. A fiducial marker assembly according to claim 51, wherein said second portion comprises one end of the housing of said imaging marker assembly.

53. A fiducial marker assembly according to claim 52, wherein said one end of the housing comprises glass.

54. A fiducial marker assembly according to claim 52, wherein said one end of the housing is a cap of the housing, said housing and cap enclosing said cavity.

55. A fiducial marker assembly according to claim 51, wherein said second portion comprises a layer near an end of said fiducial marker.

56. A fiducial marker assembly according to claim 55, wherein said second acoustic impedance is higher than said first acoustic impedance.

57. A fiducial marker assembly according to claim 51, wherein said fiducial marker is a permanent implantable fiducial marker.

58. A fiducial marker assembly according to claim 51, wherein said fiducial marker assembly is for use in an ultrasound detection technique.

59. A fiducial marker assembly according to claim 51, wherein said cavity contains one or more imaging agents visible in at least one imaging modality.

60. A fiducial marker assembly according to claim 51, wherein said first portion is said cavity and said first material is an imaging agent contained in said cavity.

61. A fiducial marker assembly according to claim 60, wherein said second portion is one end of the housing of said imaging marker assembly.

62. A fiducial marker assembly according to claim 61, wherein said one end of the housing is a cap of the housing, said housing and cap enclosing said cavity.

63. A method for automatically locating a position of a fiducial marker implanted in a bone, comprising steps of:

attaching a transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said transducer while translating it along a surface of interest near said fiducial marker;

receiving echoes from said pulsed transducer arising from a difference in an acoustic impedance of a portion of the fiducial marker and an acoustic impedance of a material located near said portion of the fiducial marker; and

determining a position in said defined coordinate system of said fiducial marker in response to said received echoes and a position and orientation of said transducer.

64. A method according to claim 65, wherein said bone is a skull of a patient and said surface of interest is a scalp of said patient.

65. A method for registering a physical space with image volumes including automatically locating a position of an object, comprising steps of.

attaching a transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said transducer while translating it along a surface of interest near said object;

receiving echoes from said pulsed transducer arising from a difference in an acoustic impedance of a portion of the object and an acoustic impedance of a material located near said portion of the object; and

determining a position in said defined coordinate system of said object in response to said received echoes and a position and orientation of said transducer.

66. A method according to claim 65, wherein said method provides said physical space to image volume registration as a tool for interactive, image-guided surgery or fractionated radiotherapy.

67. A method for automatically locating a position of an object, comprising steps of:

attaching a transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said transducer while translating it along a surface of interest near said object;

receiving echoes from said pulsed transducer arising from a difference in an acoustic impedance of a portion of the object and an acoustic impedance of a material located near said portion of the object; and

determining a position in said defined coordinate system of said object in response to said received echoes and a position and orientation of said transducer, wherein said determining step comprises steps of:

extracting a windowed surface echo signal from said received echoes;

calculating a standard deviation of said extracted windowed surface echo signal;

calculating a weighted centroid of positions of said transducer;

locating a time delay of said windowed surface echo signal nearest the weighted centroid;

calculating a depth of a center of the object in response to a depth of an interface in which said object is located and a height of said object; and

calculating coordinates in said coordinate system of the center of the object in response to said calculated depth and said time delay.

68. A method for automatically locating a position of an object, comprising steps of:

attaching a transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said transducer while translating it along a surface of interest near said object;

receiving echoes from said pulsed transducer arising from a difference in an acoustic impedance of a portion of the object and an acoustic impedance of a material located near said portion of the object; and

determining a position in said defined coordinate system of said object in response to said received echoes and a position and orientation of said transducer;

wherein said acoustic impedance of said portion of said object is lower than said acoustic impedance of said material located near said portion of said object.

69. A method for automatic amplitude-mode ultrasound location of a fiducial marker implanted in a bone, comprising steps of:

attaching an ultrasound transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said ultrasound transducer while translating it along a surface of interest near said implanted fiducial marker;

receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a portion of the implanted fiducial marker and an acoustic impedance of a material located near said portion of the implanted fiducial marker; and

determining a position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer.

70. A method according to claim 69, wherein said bone is a skull of a patient and said surface of interest is a scalp of said patient.

71. A method for automatic amplitude-mode ultrasound location of an implanted fiducial marker including a marker body and a cap together enclosing a space, comprising steps of:

attaching an ultrasound transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said ultrasound transducer while translating it along a surface of interest near said implanted fiducial marker;

receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a portion of the implanted fiducial marker

and an acoustic impedance of the cap of the implanted fiducial marker; and

determining a position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer.

72. A method according to claim 71, wherein the acoustic impedance of the cap is higher than the acoustic impedance of the portion of the implanted fiducial marker.

73. A method for registering physical space with image volumes including automatic amplitude-mode ultrasound location of an implanted fiducial marker, comprising steps of:

attaching an ultrasound transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said ultrasound transducer while translating it along a surface of interest near said implanted fiducial marker;

receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a portion of the implanted fiducial marker and an acoustic impedance of a material located near said portion of the implanted fiducial marker; and

determining a position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer.

74. A method according to claim 73, wherein said method provides said physical space to image volume registration as a tool for interactive, image-guided surgery or fractionated radiotherapy.

75. A method for automatic amplitude-mode ultrasound location of an implanted fiducial marker, comprising steps of:

attaching an ultrasound transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said ultrasound transducer while translating it along a surface of interest near said implanted fiducial marker;

receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a portion of the implanted fiducial marker and an acoustic impedance of a material located near said portion of the implanted fiducial marker; and

determining a position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer, wherein said determining step comprises steps of:

extracting a windowed surface echo signal from said received echoes;

calculating a standard deviation of said extracted windowed surface echo signal;

calculating a weighted centroid of positions of said ultrasound transducer;

locating a time delay of said windowed surface echo signal nearest the weighted centroid;

calculating a depth of a center of the implanted fiducial marker in response to a depth of an interface in which said implanted fiducial marker is located and a height of said implanted fiducial marker; and

calculating coordinates in said coordinate system of the center of the implanted fiducial marker in response to said calculated depth and said time delay.

76. A method for automatic amplitude-mode ultrasound location of an implanted fiducial marker, comprising steps of:

attaching an ultrasound transducer to a coordinate space digitizer having a defined coordinate system;

pulsing said ultrasound transducer while translating it along a surface of interest near said implanted fiducial marker;

receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a portion of the implanted fiducial marker and an acoustic impedance of a material located near said portion of the implanted fiducial marker; and

determining a position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer;

wherein said acoustic impedance of said portion of said implanted fiducial marker is lower than said acoustic impedance of said material located near said portion of said implanted fiducial marker.

77. A system for locating a position of an implanted fiducial marker, said implanted fiducial marker including a housing and a cap, said housing and cap together containing a cavity, said system comprising:

a coordinate space digitizer having a defined coordinate system;

an ultrasound transducer connected with said coordinate space digitizer;

a pulser pulsing said ultrasound transducer while translating it along a surface of interest located near said implanted fiducial marker;

a receiver receiving echoes from said pulsed ultrasound transducer arising from a difference in an acoustic impedance of a portion of the implanted fiducial marker and an acoustic impedance of the cap of the implanted fiducial marker; and

means for determining a position in said defined coordinate system of said implanted fiducial marker in response to said received echoes and a position and orientation of said ultrasound transducer.

78. A fiducial marker assembly comprising an imaging marker assembly comprising:

a portion including a material having a first acoustic impedance;

a housing; and

a marker cap, said housing and said cap together containing a cavity, said cap having a second acoustic impedance which is different than said first acoustic impedance.

79. A fiducial marker assembly according to claim 78, wherein said cap is a glass marker cap.

80. A fiducial marker assembly comprising an imaging marker assembly having a housing, said housing containing a cavity, comprising:

a first portion including a first material having a first acoustic impedance; and

a second portion including a second material having a second acoustic impedance which is different than said first acoustic impedance, said second portion comprising a layer near an end of said fiducial marker;

wherein said second acoustic impedance is higher than said first acoustic impedance.

* * * * *

UNITED STATES PATENT AND TRADEMARK OFFICE

CERTIFICATE OF CORRECTION

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DATED : March 7, 1995
INVENTOR(S) : Judith T. Lewis et al

It is certified that error appears in the above-indentified patent and that said Letters Patent is hereby corrected as shown below:

Column 13, line 23, change "image-guide" to --image-guided--.

Column 15, line 12, change "claim 29" to --claim 32--.

Column 17, line 32, change "claim 65" to --claim 63--.

Signed and Sealed this

Twentieth Day of June, 1995

Attest:

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Attesting Officer

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